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THE BIOLOGICAL EFFECTS OF VIBRATION

D. E. Goldman, Editor

Armed Forces - NRC Committee on Hearing and Bio-Acoustics

Working Group 39

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ARMED FORCES – NATIONAL RESEARCH COUNCIL
COMMITTEE ON HEARING AND BIO-ACOUSTICS

THE BIOLOGICAL EFFECTS OF VIBRATION

Report of Working Group 39

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Office of Naval Research
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CONTENTS

THE BIOLOGICAL EFFECTS OF VIBRATION

SUMMARY

This report discusses the effects of vibration on man. It summarizes briefly: (a) the measurement of vibration, (b) the production of controlled vibratory stimulation for research purposes, (c) the injurious effects of vibration, (d) discomfort due to vibration, (e) effects of vibration on task performance, and (f) beneficial uses of vibration.

On the basis of present knowledge, it may be stated that:

1. The accurate measurement of vibration is feasible in most situations.

2. The controlled production of vibratory stimulation is practicable over a fairly wide range of conditions.

3. Most biological effects of interest (excluding motion sickness and acoustic phenomena) occur in the frequency range, 2 to 20 cycles per second (cps), and at peak accelerations above 0.001 g.

4. The biological effects of vibration depend not only on frequency and amplitude of vibration, but also on the duration and mode of exposure.

5. It is not now possible to establish definite safety or comfort limits, or to specify effects on task performance. Ranges of importance can, however, be indicated.

   a. Peak accelerations above about 1 g may be dangerous under certain conditions.

   b. Peak accelerations above about 0.03 g may be disturbing in one way or another.

6. More extensive research needs to be done to permit more useful generalizations. The following areas of research are of particular importance:

   a. Examination of effects of non-sinusoidal vibration.

   b. Measurement of effects of vibration upon task performance.

   c. Accumulation of information on vibrations actually occurring in vehicles.

   d. Correlation of studies on vibration with work on shock, impact, and other kinds of mechanical force.

7. There is an important need for establishing a means by which information can be more readily exchanged between engineers and operating personnel on the one hand, and those concerned with health and safety on the other.

8. Research on the biological effects of vibration is expensive because of (a) the costs of equipment to produce and measure vibration, and (b) the time required to conduct many experiments. Personnel with appropriate scientific training are difficult to obtain.

RECOMMENDATIONS

1. Continuing support at a relatively high priority should be given to those laboratories—military, industrial, and university— which have the facilities and trained personnel for research on the biological effects of vibration.

2. The collaborative efforts of engineers, biophysicists, psychologists, and physicians are needed to establish more accurately the tolerance limits for injury due to vibration. Particular attention should be given to the effects of random vibration (mechanical noise).

3. Encouragement should be given to studies of the effects of vibration on different kinds of task performance as well as on discomfort and fatigue. This is an exceedingly difficult problem since results may vary greatly with the individual, the kind of task performed, the environment in which measurements are made, and numerous other factors. Nevertheless, attempts should be made to obtain data which will permit rough generalizations.

4. More extensive measurements should be obtained on the vibratory motions which occur in vehicles or in the use of equipment, especially where large forces and energies are involved.

5. The possible beneficial effects of vibration need to be more carefully examined.

6. Research on the use of vibratory stimuli for communication should be continued.

7. An agency of some kind should be established to serve as a clearing house for the exchange of information on vibration, shock, etc., as it affects man and as it concerns the design, construction, and use of vehicles and machines.

REPORT

Working group 39 was established by CHABA in November 1959 in response to requests for an evaluation of the present state of knowledge of the effects of vibration on man. It has been evident for several years that the mechanical forces applied to people in or near vehicles and certain other machines were becoming more and more a matter of concern. Outright injury from exposure to vibratory forces has been rare except for localized injuries resulting from the handling of vibrating tools over a long period of time. Difficulties due to mechanical interference with task performance have sometimes arisen in connection with the operation of vehicles. Discomfort and fatigue from long exposure to vibration seem to be relatively common and, while these are not by themselves dangerous, they can contribute to the lowering of the effectiveness of vehicle operators and thus take their place among the causes of accidents. The accelerating pace of design and construction of high performance machines makes it increasingly important that design criteria be made available which will reduce the necessity for expensive modifications. This report undertakes to outline very briefly what is now known about the effects of vibration on man and what can and should be done to extend useful knowledge of the subject. A detailed discussion of several aspects of the problem will be found in the Appendix.

Vibration first became an industrial hygiene problem in the 19th century when pneumatic drills were introduced. With the development of industry and of transportation, the problem has grown steadily. A number of years ago, several studies were made to establish limits above which discomfort might be produced. The difficulty of carrying out observations of this kind and the lack of sophistication in many of the experiments make the data rather crude. Nevertheless, little has been added, and the best estimates now obtainable on thresholds for perception and for discomfort rest largely on this early work.

Concern with the problems of "tolerance" limits and injury has introduced another problem. The intensity of vibration required to produce discomfort is not great, but if one wishes to study injury, relatively heavy and carefully designed machinery is required. Such apparatus is expensive and requires a substantial and well-trained maintenance staff. There have been few places where this kind of work can be done properly.

Definitive studies on the effects of vibration require the services of people trained in several disciplines: engineering, biophysics, psychology, and medicine. Much of the work on high intensity vibration must be done with animals, thus producing difficulties in the interpretation of the results. Up to the present time, nearly all studies have been made using sinusoidal vibration. Current interest, however, centers about random "mechanical noise" occurring in relatively high-speed travel in an irregular environment. Included in this area of interest are aircraft in high speed, low altitude flight, small vessels in rough water, heavy vehicles on rough terrain, and projected conditions in space vehicles.

A definition of mechanical vibration is in order here. For the present purposes it is to be understood as referring to alternating motion in a frequency range such that the system under observation behaves more as if it were composed of discrete elements than as a system in which wave motion predominates. For a human body this is true for frequencies below about 20 cps. On the other hand, this report is not concerned with motion sickness and, therefore, frequencies below about 2 cps are excluded from consideration.

In order to assess the possibilities of understanding the phenomena it is convenient to raise and discuss a few specific questions:

1. What is the present situation with respect to the measurement of vibratory forces to which vehicle occupants may be exposed?

Engineering techniques in this field are rather well developed and the difficulties encountered in obtaining useful data arise primarily from considerations of cost, time, and interest. Measurement of the vibration of certain parts of the body may, however, be quite difficult.

2. What are the basic factors determining and mediating the deleterious effects of vibration on man?

The body is a complicated mechanical structure, and it is necessary to know something about its dynamic mechanical properties in order to understand the way in which mechanical responses occur. There are now available a number of measurements of mechanical impedance, resonance, and damping characteristics of the body for several modes of excitation. The frequency range in which most significant mechanical responses occur is approximately from 2 to 20 cps. Since the human body acts as a pure mass, below about 2 cps, the problem is primarily one of vestibular reactions. At frequencies above 20 cps, the problems become more and more the province of acoustics. Mechanical injury is related to relative displacements of body organs, and the limited material which has been obtained on severe disturbances and injuries has been interpreted in terms of the mechanical properties of the body. Biological responses, i.e., those in which the body takes an active part, are less well worked out. Stimulation of sensory systems produces reflex and higher nervous responses which may affect task performance significantly and contribute to discomfort and fatigue. There may also be generalized "stress responses."

3. To what extent is it now possible to establish criteria for comfort and safety?

Studies made up to the present time indicate that one can point to ranges of frequency and acceleration in which certain types of responses may be expected to appear. In rough terms, and disregarding resonance effects, one can say that in the range from 2 to 20 cps vibration becomes detectable by the human body at accelerations somewhat above 0.001 g and may become dangerous at accelerations above about 1 g. Certain intermediate response levels can also be indicated approximately. Very little is known about specific behavioral responses in particular task situations and no general statements can now be made on this subject.

4. What further work needs to be done to provide a better understanding of the effects of vibration on man?

From what has already been said, it is clear that while the effects of sinusoidal vibration have been studied to some extent, the problem of "mechanical noise" needs special attention. This is not only because of immediate practical problems but because the nonlinearity of biological systems makes extrapolation from sinusoidal forces to random forces unreliable.

Problems of task performance are of particular importance. It is difficult, perhaps impossible, to apply results obtained for one specific task to another specific task. However, the need for extensive data is increasing, and emphasis must be placed on the care with which such studies are planned and carried out. The fact that these studies are expensive and time-consuming makes a casual approach wasteful.

Accumulation of information on vibratory motion actually occurring in vehicles is also needed. At the present time, little seems to be available and more should be obtained. Again, the cost and time elements are important. Continuing effort should be placed on enlarging the scope of experimental material already at hand in order to improve its generality and accuracy. It is too much to expect that precise numerical values can be set as limits for comfort or safety, but as more extensive data are collected, detailed knowledge of the variability of human subjects and of exposure conditions should permit better assessment of relative risks and, consequently, of relevant protective measures.

Attention should also be called to the close relationship of vibration to shock, impact, and other types of mechanical force in that the mechanical properties of human beings underlie responses to all kinds of mechanical force. Much of what is learned from studies on one type of force is of value in interpreting material arising out of studies on another.

BENEFICIAL USES OF VIBRATION

A problem which has not yet been referred to is that of the possible value of vibration as a useful stimulus. This includes the use of vibration to help counteract sensory isolation, to help maintain superficial circulation and muscle tone, and as a route for communication. Very little is known about these possibilities, although the last mentioned has been the subject of some research activity and has been discussed at an earlier CHABA meeting. These matters need to be looked into in some detail because of their potential importance.

LIAISON

Since so many problems relating to the effects of both shock and vibration in man arise from engineering and operational developments, it is important that close liaison be maintained between those concerned with the design, construction, and use of machinery and vehicles on the one hand, and those concerned with health and safety on the other. No effective liaison mechanism now exists. It is therefore strongly recommended that CHABA take the initiative in establishing such a mechanism either as part of its own operations or elsewhere. There is a real need for some kind of informal group to act as a clearing house for information obtained and needed by the Armed Forces, industry, and other organizations. The fact that most work in this field requires expensive apparatus and a special staff points definitely to the value of mutual consultation. Furthermore, those agencies concerned with the setting of criteria must be made aware of what is available and what is being done. In a field which is growing and shifting as this one is, such coordination is specially important.

APPENDIX

Please note that the numbering of pages in the Appendix differs from that of the report since the Appendix is actually a discrete report which has been released for use by CHABA through the courtesy of the Naval Medical Research Institute, and as such has its own system of pagination.

THE EFFECTS OF SHOCK AND VIBRATION ON MAN

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Issued by the Naval Medical Research Institute.

INTRODUCTION

This review deals with three problems: 1) the determination of the structure and properties of the human body considered as a mechanical as well as a biological system, 2) the effects of shock and vibration forces on this system, and 3) the protection required by the system under various exposure conditions and the means by which this protection is to be achieved. Man, as a mechanical system, is extremely complex and his mechanical properties are quite labile. He is also a human being and in this capacity refuses to permit destructive testing but will nevertheless expose himself carelessly to mechanical damage which may arise from situations of his own making and will then demand a degree of protection against this damage which shall be almost impossibly effective, unhampering, and cheap. Because of such conflicting attitudes there is very little reliable information on the magnitude of the forces required to produce mechanical damage to people. It is therefore necessary to use experimental animals for most studies on mechanical injury. However, the data so obtained must be subjected to careful scrutiny to determine the degree of their applicability to humans, who differ from animals not only in size but in anatomical and physiological structure as well. It is only occasionally possible to obtain useful information from situations involving accidental injuries to man, since while the damage can often be assessed, the forces producing the damage cannot. It is also very difficult to obtain reliable data on the effects of mechanical forces on the performance of various tasks and on subjective responses to these forces largely because of the wide variation in the human being in both physical and behavioral respects. Measurement of some of the mechanical properties of man is, however, often practicable since only small forces are needed for such work. The difficulty here is in the variability and lability of the system and in the inaccessibility of certain structures.

DEFINITIONS AND CHARACTERIZATION OF FORCES

Characterization of Forces. - In relation to effects on human beings it is convenient to classify mechanical forces as steady, impulsive or alternating and to distinguish accelerative forces and ambient pressures which act on the entire body from localized forces. The effects of steady forces are not discussed here nor is it possible to consider in detail the tremendous number of ways in which the various kinds of mechanical force can be applied to the human body. These forces may be transmitted to the body through gas, liquid, or solid. They may be diffused or concentrated over a small area. They may vary from tangential to normal in more than one direction. The shape of a solid body impinging on the surface of the human is as important as the position or shape of the human body itself. All of these factors must be taken into account in comparing injuries produced by vehicle crashes, explosions, blows, vibration, etc. Laboratory studies often permit fairly accurate control of force application but field situations are apt to be extremely complex and thus make it very difficult to transfer information between the laboratory and the field.

Shock. - The term "shock" basically refers to an event which is sufficiently rapid and of great enough magnitude to be significant. However, the notion has many extensions and the usage has developed differently in engineering than in biology and medicine; therefore one must be careful not to confuse the various meanings given to the term. Hemorrhagic shock, for example, may result from mechanical shock but the two are not the same. The term "shock" is used in this review in its engineering sense. A *shock wave* is a discontinuous pressure change propagated through a medium at a velocity greater than that of sound in the medium. A blast wave may or may not have a discontinuity. In any case, rates of change of applied forces are considered fast or slow primarily with reference to those time constants of an affected system which are of interest. As a rule, forces reaching peak values in less than a few tenths of a second and of not more than a few seconds duration are shock forces in relation to the human system.

The terms *impact* and *blow* refer to forces applied when the human body comes into sudden contact with a solid body and when the momentum transfer is considerable as in rapid deceleration in a vehicle crash or when a rapidly moving solid body strikes a human body.

Vibration. - Biological systems may be influenced by vibration of sufficient amplitudes at all frequencies. This review is concerned primarily with the frequency range from about 1 c.p.s. to several hundred c.p.s. although studies at higher frequencies are very useful for the analysis of tissue characteristics.

Mechanical noise and random shocks form a group intermediate between shock and vibration. They may often be treated either as a vibration spectrum or as repeated individual shocks. The relationship between the repetition rates and the resonance characteristics of the system acted upon is important in determining the approach.

METHODS AND INSTRUMENTATION

Most quantitative investigations of the effects of shock and vibration on humans have been conducted in the laboratory in controlled, simulated environments. Meaningful results of such tests can be obtained only if measurement methods and instrumentation are adapted to the particular properties of the biological system under investigation so that noninterference of the measurement as such with the system's behavior is assured. This behavior may be physical, physiological, and psychological although these parameters should be studied separately if possible. The complexity of a living organism makes such separation, even assuming independent parameters, only an approximation at best. In many cases if extreme care is not exercised in planning and conducting the experiment, uncontrolled interaction between these parameters can lead to completely erroneous results. For example, the dynamic elasticity of tissue of a certain area of the body may depend on the simultaneous vibration excitation of other parts of the body, or it may change with the duration of the measurement since the subject's physiological response varies, or it may be influenced by the subject's psychological reaction to the test or the measuring equipment (startle, fright, etc.). The design of an experiment must consider all these factors and select carefully the most promising approach.

Control of, and compensation for, the nonuniformity of living systems is absolutely essential. People vary in size, shape, sensitivity, and responsiveness, and these factors for a single subject vary with time, experience and motivation. The use of adequate statistical experimental design is necessary and almost always requires a large number of observations and carefully arranged controls.

PHYSICAL MEASUREMENTS

The mechanical force environment to which the human body is exposed during physical measurements must be clearly defined. Force and vibration amplitude should be specified for the area of contact with the body. Vibration measurements of the body's response should be made whenever possible—which is unfortunately not often—by non-contact methods. X-ray methods have been used successfully to measure the displacement of internal organs. Optical, cinematographic, and stroboscopic observation can give the displacement amplitudes of larger parts of the body. Small vibrations can sometimes be measured without contact by capacitive probes located at small distance from the (grounded) body surface. The probe forms a condenser with the body; capacity changes due to vibrations can be measured in a high-frequency carrier system. If vibration pickups in contact with the body are used, they must be small and light enough so as not to introduce a distorting mechanical load. This usually places a weight limitation on the pickup of a few grams or less, depending on the frequency range of interest and the effective mass to which the pickup is attached. Figure 1 illustrates the effect of weight and size on the response of accelerometers attached to the skin overlying soft tissue (1). The lack of rigidity of the human body as a supporting structure makes measurements of acceleration usually preferable to those of velocity or displacement. The mechanical impedance of a sitting, standing, or supine subject is extremely valuable for calculating the vibratory energy transmitted to the body by the vibrating structure (2). The ratio and phase angle of driving force to resulting velocity can be measured for this purpose in many different ways. Care must be taken to ensure that the measurement of the total force transmitted to the body is not influenced by load distribution. The impedance of small areas of the body surface has been measured with vibrating pistons (3), resonating rods, or acoustical impedance tubes (4).

If the whole body is exposed to a pressure or blast wave in air or water, exact definition of the pressure environment is essential. Calculation of the pressure distribution over the body (amplitude and phase) and the pressure increase at the body produced by an undistorted free-field pressure is difficult; for blast waves it is almost impossible. Therefore the pressure distortion should be measured if possible (5). Orientation

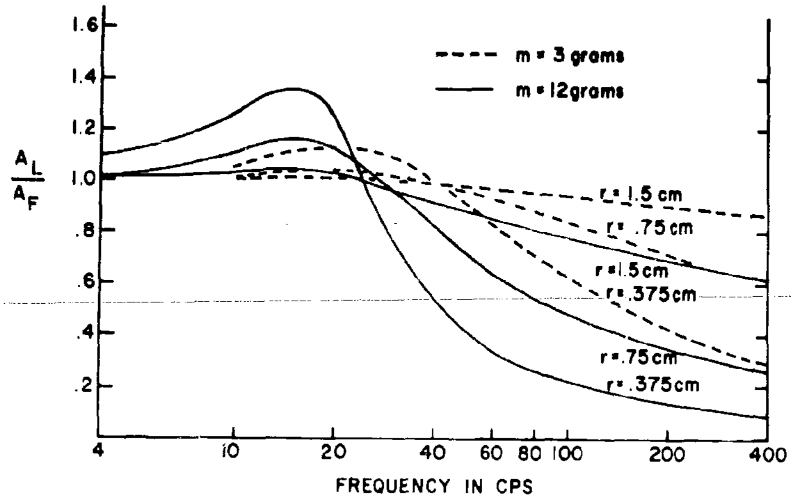


Figure 1. - Amplitude distortion due to size and weight of accelerometer attached to body surface over soft tissue of human subject exposed to vibration. The graph gives the ratio, $\frac{A_L}{A_F}$, of the response of a loaded accelerometer of mass m and radius r of contact surface area for several values of m and r. [Values calculated from mechanical surface impedance data of Franke and von Gierke (1).]

of the subject with respect to the wave front must be known. Dynamic pressure or "windage" must be considered. If the environment deviates from free-field conditions it should be carefully specified because of its effect on peak pressure and pressure time history (6).

BIOLOGICAL MEASUREMENTS

Physiological instrumentation for such characteristics as blood pressure, respiration rate or depth, heart potential, brain potential, or galvanic skin response—which has been used as a quantitative indicator for the over-all vibration load (7)—must be carefully checked for freedom from artifacts when the subject, the instrument, or both are exposed to vibration, intense sound, or acceleration. Conventional commercial instruments and clinical methods of using them are frequently unsatisfactory for such tests. A bio-electronic harness (8) such as is used in aviation medicine is very useful for measurement of various physiological functions and lends itself to simple arrangements in field tests, but proper functioning of instruments must be verified for each environment.

If psychological experiments during exposure to the mechanical stimuli are planned, adherence to established rules for such subjective tests is an absolute necessity for valid results. The maintaining of a neutral situation with uniform motivation, subject instruction, and adequate statistical design of the experiment are among the most important considerations. Care must be taken that the subject be not biased or influenced by environmental factors not purposely included in the test (e.g., the noise of a vibration table can act as a disturbing factor in a study of vibration effects). Subjective observations of physical phenomena (made as subject or as "observer") can be distorted by many factors and should not be allowed to compete with physical measurements.

SIMULATION OF MECHANICAL ENVIRONMENT

The desire to study the physical, physiological, and psychological responses of biological specimens in the laboratory, under well-controlled conditions, has led to the use of standard and specialized shock and vibration testing machines. A summary of some machines employed in such tests is given in figure 2. As in the case of the environmental testing of physical components and systems, an accurate simulation of man's actual mechanical environment is frequently not feasible for technical or economic reasons or may even be undesirable because of a need for more systematic investigation under somewhat simplified conditions. Thus most investigations are limited to the study of one degree of freedom at a time and many fundamental studies are performed with sinusoidal forces. Adequate safety precautions, safe and accurate control of the exposure, and sufficient load capacity for subject, seat, and instrumentation are obvious requirements for all shock and vibration machines used. Mechanical and electro-dynamic shake tables are employed for this purpose. Since the law of linear superposition is only valid in the linear phsyical domain,

TYPE OF MACHINE	APPLICATION OF FORCE	FORCE-TIME FUNCTION	FREQUENCY RANGE	MAXIMUM LOAD IN EXPERIMENTS	REFERENCE (APPLICATION IN BIOLOGY EXPERIMENT)
SHAKE TABLE			MECHANICAL 0-50 cps	UP TO 15g	10 2
			ELECTRODYNAMIC 15-1000 cps		
VERTICAL ACCELERATOR			0-10 cps	± 10 FT 3.7g PEAK	11
SHOCK MACHINE			DOWN TO T = 16 sec τ = 2 10^{-3} sec	PEAK AMPLITUDE 10^{-2} TO 10^{-1} cm	12
HORIZONTAL OR VERTICAL DECELERATOR OR ACCELERATOR (SLEDS ON TRACKS, CAR, DROP-TOWER)			RATE OF ACCELERATION UP TO 1400 g/sec	40 g PEAK	9 13 14 15
BLAST TUBE FIELD EXPLOSION				—	11, 18 5 17
SIREN (AIRBORNE SOUND)			25-100000 cps	160-170 db re 0.0002 μbar	
RESPIRATOR			0-15 cps	—	19
VIBRATOR (SMALL PISTON)			0-10 Mcps	-	20 5 21
SHAKER ON CENTRIFUGE	ALTERNATING g FORCE IN ADDITION TO STATIC g FIELD		0-3 cps	—	22
HEAD IMPACT MACHINE FOR DUMMY HEADS			DEPENDING ON STRUCTURE STRUCK	IMPACT VELOCITY 140 FT/SEC	23

Figure 2. - Summary of characteristics of shock and vibration machines used for human and animal experiments. Force time functions are indicated schematically. Frequency range and maximum load refer to values used, not to capabilities of machines. References are to papers describing use of the machines for biological purposes.

sinusoidal forces alone are not adequate for the study of non-linear physical responses or physiological and psychological reactions to complex force functions. Some of the machines listed are specially designed for exposure of humans. The vertical accelerator, for example, employs a friction drive mechanism to permit the simulation of large amplitude sinusoidal and random vibrations such as those encountered in buffeting during low altitude high speed flight or anticipated during the launch and re-entry phases of spacecraft (11). The device can be programmed within its performance limits with acceleration recordings obtained under actual flight conditions. Upward and downward acceleration tracks have also been built with sliding seats projected by explosive charges to study human tolerance to ejection from high-speed aircraft (ejection seat). Horizontal tracks with rocket propelled sleds, which can be stopped by braking mechanisms with variable functions, have been used to study the effects of linear decelerations similar to those occurring in automobile or aircraft crashes (9). Interest has recently developed in the study of combinations of increased static acceleration with transient accelerations and vibrations. This is done by mounting vibrators on centrifuges. Blast tubes, sirens, and body respirators are used to study the body's response to pressure distributions surrounding it. At low frequencies the respirator has been valuable in studying the response of the lung-thorax system. Small vibrating pistons, which are available for a wide frequency range, have been used in investigating the mechanical impedance of small surfaces, the transmission of vibration and the physiological reaction to localized excitation. The range of some high acceleration devices is shown in figure 3.

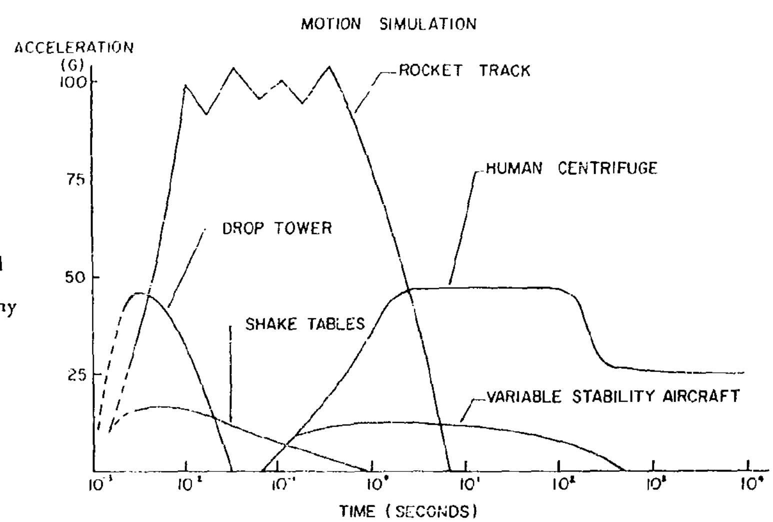


Figure 3. - Ranges of time and acceleration obtainable with certain devices [after Clark and Gray (24)].

SIMULATION OF HUMAN SUBJECTS

The establishment of human tolerance limits to mechanical forces and the explanation of injuries produced when these limits are exceeded frequently requires experimentation at various degrees of potential hazard. To avoid unnecessary risks to humans, different types of animals are first used for detailed physiological studies. As a result of these studies, levels may be determined which are, with reasonable probability, safe for human subjects. However, the obvious limitations of such comparative experiments must be kept in mind. The different structure, size, and weight of most animals shifts their response curves to mechanical forces into other frequency ranges and to other levels than those observed on humans. This must be considered in addition to the general and partially known physiological differences between species. For example the natural frequency of the thorax-abdomen system of a human subject is between 3 and 4 c.p.s.; for a mouse the same resonance occurs between 18 and 25 c.p.s. Maximum effect and maximum damage therefore occur at different vibration frequencies and different shock-time patterns than in the human case. However, the principal laws of response and effect are well worth studying on small animals if care is taken in the interpretation of the data and if scaling laws are established. Dogs, pigs, and monkeys are used extensively in such tests.

Many kinematic processes, physical loadings, and gross destructive anatomical effects can be studied on dummies which approximate a human being in size, form, mobility, total weight, and weight distribution in body segments (25). Several such dummies are commercially available. In contrast to those used only for load purposes, dummies simulating basic static and dynamic properties of the human body are called "anthropometric" or "anthropomorphic" dummies. They have been used extensively in aviation and automotive crash research and in other studies to precede work with human subjects and to study protective seats and harnesses. Such dummies attempt to match the "resiliency" of human flesh by some kind of padding, but they are crude simulations at best and the dynamic mechanical properties are, if at all, only reasonably matched in a very narrow low-frequency range. This and the passiveness of such dummies must be kept in mind as important mechanical differences between them and living subjects.

Efforts have been made to simulate the mechanical properties of the human head more closely in order to study the physical phenomena occurring in the brain during crash conditions (23). Although these forms only approximate the human head, they are very useful in the evaluation of the protective features of crash, safety, and anti-buffet-helmets. Plastic head forms, conforming to standard head measurements, are designed to fracture in the same energy range as that established for the human head. A cranial vault is provided to house instrumentation and a simulated brain mass with comparable weight and consistency (a mixture of glycerine, ethylene glycol, etc.). The static properties of the skin and scalp tissue have been simulated with poly-vinyl foam.

The static and dynamic breaking strength of bones, ligaments, and muscles and the forces producing fractures in rapid decelerations have frequently been studied on cadaver material (26,27,28). Extreme caution must be exercised in applying elastic and strength properties obtained in this way to a situation involving the living organism. The differences observed between wet, dry, and embalmed preparations are considerable (29) and changes in these properties will also result in changes in the force distribution on a composite structure.

A multitude of physiological, anatomical, and physical factors must be considered for each specific situation, in which the use of animals, dummies, or cadavers as substitutes for live human subjects is planned. Only such careful analysis paired with long experience will make either of these methods a worth-while tool for prediction and extrapolation.

PHYSICAL CHARACTERISTICS

ANATOMY

Structurally, the body consists of a hard, bony skeleton whose pieces are held together by tough, fibrous ligaments and which is embedded in a highly organized mass of connective tissue and muscle. The soft visceral organs are contained within the rib cage and the abdominal cavity.

An outline of the skeleton is shown in figure 4. The slightly bowed vertebral column is the central structural element. It consists of a number of individual vertebrae which have roughly cylindrical load-bearing elements separated by fibro-cartilaginous pads. Near the lower end several vertebrae are fused together to form the sacrum which is a tightly fitting part of the pelvic girdle. The skull rests at the top of the column and is held in place by muscle and connective tissue as well as by ligaments. At the bottom of each side of the pelvic girdle is a roughly hemispherical hollow into which the head of the femur fits. Below the femur are the tibia and fibula which in turn rest on the ankle and foot bones.

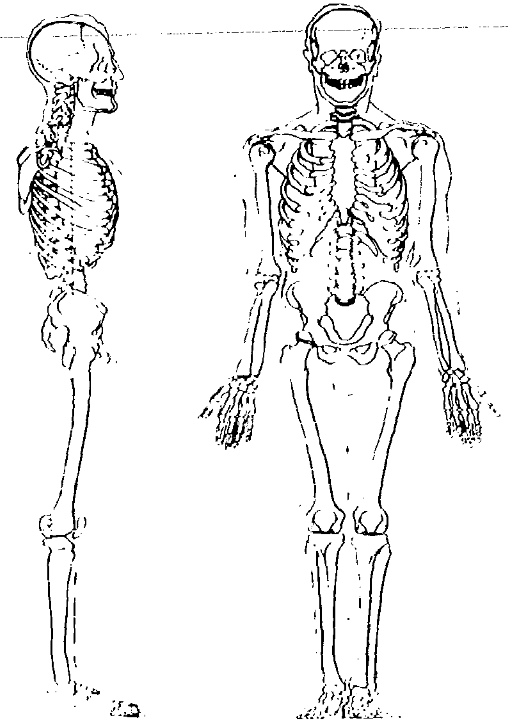


Figure 4. - Diagram of human skeleton [from Goldman (34)].

The intervertebral discs are dense fibro-cartilaginous pads. The hip, knee, and ankle joints have cartilaginous layers on their articular surfaces as do also the joints of the upper limbs. The foot has a tough connective tissue pad at the heel and an arrangement of bones which acts to distribute applied loads. All the joints are held together by ligaments which are flexible but relatively inextensible. These ligaments form a more or less crisscross lacing which permits movement of the joints in a suitable direction without stretching the ligaments themselves to an appreciable extent. The sacroiliac joint is held tightly and almost immovably. The rib cage and shoulder girdle are also dependent on muscle and connective tissue for their support.

In the ideal standing position, a plumb line through the center of gravity of the body passes through the lower lumbar and upper sacral vertebrae, slightly behind the hip joint sockets and a bit in front of the knee and ankle joints. Upward, the line passes in front of the thoracic curve of the vertebral column and through the support at the base of the skull. Vertical thrusts may be taken up by the compression of the joint pads and by bending of the vertebral column. There is often a slight forward turning moment at the pelvis especially in older adults. Small deviations from postural symmetry may result in a markedly asymmetrical force distribution under the action of vertical thrusts.

The body musculature, supported from the skeleton by tendons and held together by a network of connective tissue fibers, forms a secondary supporting structure for the skeleton and joints. Fat and skin also contain connective tissue.

In compression, soft tissues resemble water in their mechanical properties, but in shear they approximate stiff, nonlinear gels with internal losses.

The visceral organs (fig. 5) contained within the thoracic cage and abdominal cavity, are soft tissue elements, separately encapsulated to slide freely over each other, and supported individually by membranes and ligaments and collectively by the bone, muscle, and connective tissue surroundings. Their weights range from a fraction of an ounce to several pounds and most of the supporting membranes are very flexible. The kidneys are embedded in fatty tissue and held by a connective tissue sheath in a depression in the posterior wall of the abdominal cavity. The stomach is supported by the esophagus and its displacement is restricted by the dome-shaped diaphragm which is a large sheet of muscle separating the chest cavity from the abdominal cavity. The lungs, filled with tiny air sacs, are held in place by a combination of supports including a pressure differential. The heart is held in place by ligaments stretched longitudinally through the chest cavity and by large blood vessels. Considerable support is also provided by the diaphragm.

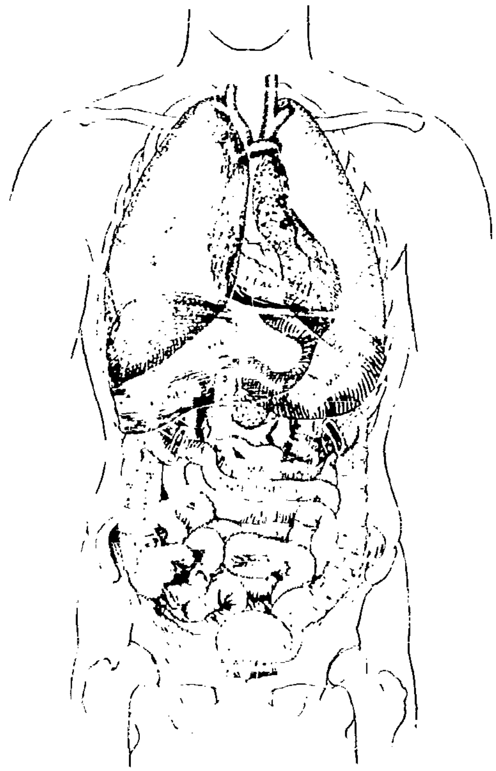


Figure 5. - Diagram of position of human viscera [from Goldman (34)].

The brain and spinal cord have special protection. The former is surrounded by liquid contained mostly in the spongy sub-arachnoid space inside the skull. The spinal cord runs longitudinally through holes in the vertebrae lined with heavy ligaments and a membrane which forms a liquid-filled tube.

Fluid in the body consists of 1) 5 to 6 liters of blood in the heart, arteries, veins, and capillaries; 2) 100 to 150 $cm.^3$ of cerebrospinal fluid surrounding the brain and spinal cord and also with the ventricular cavities of the brain; 3) the interstitial fluid found everywhere in the body as a bathing fluid for cells and tissues but nowhere in large reservoirs; and 4) liquid contained from time to time in the stomach, intestines, and bladder. Gas occurs in the sinuses of the skull, the oronasal cavity, the trachea, the lungs, and often in the stomach and intestines The latter two organs often contain solid matter as well.

Table 1. Mechanical Characteristics of the Human Body and Some of Their Applications

Dynamic Mechanical Quantity Investigated	*Problem Area of Application*
Skull resonances and viscosity of brain tissue	**Head injuries; bone-conduction hearing**
Impedance of skull and mastoid	**Matching and calibration of bone-conduction transducers; ear protection**
Ultrasound transmission through skull and brain tissue	**Brain tumor diagnosis; changes in central nervous system exposed to focused ultrasound**
Sound transmission through skull and tissue	**Bone-conduction hearing**
Mechanical properties of outer, middle and inner ear	**Theory of hearing; correction of hearing deficiencies**
Resonances of mouth, nasal and pharyngeal cavities	**Theory of speech generation; correction of speech deficiencies; oxygen mask design**
Resonance of lower jaw	**Bone-conduction hearing**
Response of mouth-thorax system	**Blast wave injury; respirators**
Propagation of pulse cardiac pressure	**Circulatory physiology; hemodynamics**
Heart sounds	**Physiology of heart; diagnosis**
Suspension of heart	**Ballistocardiography; injury from severe vibrations and crash**
Response of the thorax-abdominal mass system	Severe vibration and crash injury; crash protection
Impedance of subject sitting, standing or lying on vibrating platform	Isolation and protection against vibration and short time accelerations; ballistocardiography
Impedance of body surface, surface wave velocity, sound velocity in tissue, absorption coefficient of body surface	Theory of energy transmission and attenuation in tissue; determination of tissue elasticity, viscosity and compressibility; determination of acoustic and vibratory energy entering the body, vibration isolation; design of vibration pickups; transfer of vibratory energy to inner organs and sensory receptors
Absorption of ultrasound in tissue	Theory of energy transmission on cellular basis determination of doses for ultrasonic therapy.

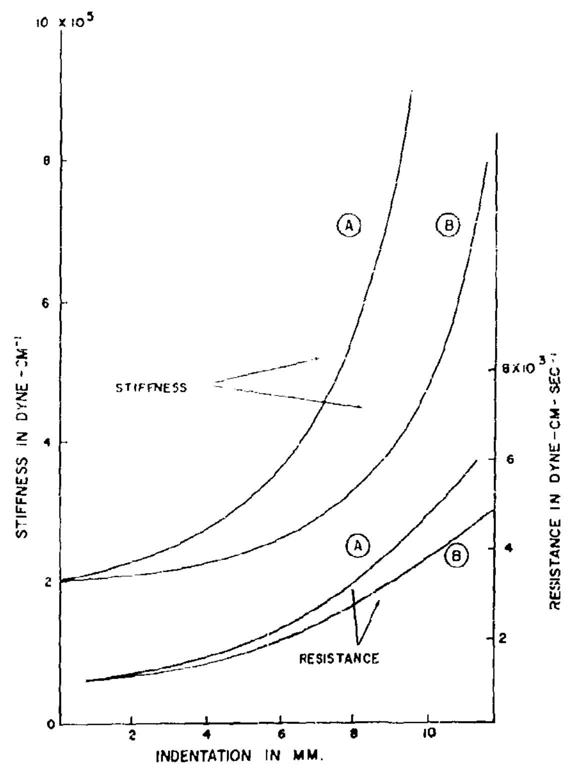


Figure 6. - Effect of loading distortion of body surface on surface impedance of soft tissue for two experimental human subjects A and B [after Franke (21)].

The cells of which the body is made have a wide variety of size and shape. Diameters range from about 0.0001 to 0.01 inch. Shapes may be spherical, disclike, columnar, flat or highly irregular. Many cells have filamentous processes projecting from them. The internal structure of cells is also very complex. They contain salts, protein, carbohydrates, and many other substances. Approximately 60 to 80 percent of the cell is water. Nuclei and other inclusion bodies are found. The rest of the cell is a viscous solution or gel with evidence of considerable submicroscopic structure.

Soft tissues consist of cells held together by connective tissue and by intercellular links. Blood is a liquid containing nearly 50 percent by volume of disclike red cells together with a few white cells. Soft tissues exhibit a wide variety of structures. Striped (voluntary) muscle consists of parallel bundles of long thin cells which can be either relaxed or contracted. Control of contraction is provided by nerve fibers which act on small groups of muscle fibers, and the elastic stiffness of the whole muscle can vary widely. Smooth (involuntary) muscle occurs mostly in the walls of hollow organs such as the stomach, intestines, blood vessels, and other specialized organs. The heart consists of a specialized type of muscle fiber.

Nerve tissue is partly cellular (grey matter) and partly fibrous (white matter). The latter contains considerable fatty material in the fiber sheaths.

Bone is also very complex. There is an outer layer of hard compact material underneath which is a layer of looser, spongelike bone so arranged as to produce a maximum of strength for commonly encountered stresses. The marrow of some bones contains blood-forming tissue.

The density of most soft tissue is between 1.0 and 1.2 with fatty tissue being somewhat lighter and bone somewhat heavier. Lung tissue is lighter still because of its air content.

PHYSICAL CONSTANTS AND MECHANICAL TRANSMISSION CHARACTERISTICS

Use of the Physical Data. - This section summarizes briefly what is known about the passive mechanical responses of the human body and tissues exposed to vibration and impact. The data can be used to calculate quantitatively the transmission and dissipation of vibratory energy in human body tissue, to estimate vibration amplitudes and pressures at different locations of the body and to predict the effectiveness of various protective measures. Frequently data can be applied to many problems other than the one for which the original study was undertaken. Table 1 summarizes some kinds of dynamic mechanical characteristics which have been studied and indicates some of the areas of application. In cases where detailed quantitative investigations are lacking the information may still serve as a guide for the explanation of observed phenomena or for the prediction of results to be expected. Most physical characteristics of the human body presented in this paragraph, except for the strength data, have been obtained by assuming the body to be a linear, passive mechanical system. This is an idealization which holds only for very small amplitudes and must be kept in mind if mechanical injury to tissue is considered. There is actually considerable nonlinearity of response well below amplitudes required for the production of damage. An example of this is given in figure 6. Whereas bone behaves more or less like a normal solid, soft elastic tissues like muscle, tendon, and connective tissue resemble elastomers with respect to Young's modulus, S-shaped stress-strain relation and large stretchability. Their properties have been studied in connection withthe quasi-static pressure-volume relations of hollow organs such as arteries, the heart, the urinary bladder, and so on (30), but linear properties have always been assumed when dynamic responses were studied. Soft tissue can then be described phenomenologically as a visco-elastic medium and plastic deformation has to be considered only if injury occurs. Approximate physical properties representative of human body tissue are summarized in table 2.

The combined use of soft tissue and bone in the structure of the body together with the body's geometric dimensions results in a system which exhibits roughly three different types of response to vibratory energy depending on the frequency range: at very low frequencies, below approximately 100 c.p.s., the body can be

Table 2. Physical Properties of Human Tissue

	Tissue, Soft	*Bone, Compact*	
		Fresh	*Embalmed, Dry*
Density, g/cm.^{3}	1-1.2	1.93 – 1.98	1.87
Young's Modulus, dyne/cm.^{2}	7.5×10^4	2.26×10^{11}	1.84×10^{11}
Volume compressibility,* dyne/cm.^{2}	2.6×10^{10}	–	1.3×10^{11}
Shear elasticity,* dyne/cm.^{2}	2.5×10^4	–	7.1×10^{10}
Shear viscosity,* dyne sec/cm.^{2}	1.5×10^2	–	–
Sound velocity, cm/sec.	$1.5\text{-}1.6 \times 10^5$	3.36×10^5	–
Acoustic impedance, dyne sec/cm.^{3}	1.7×10^5	6×10^5	6×10^5
Tensile strength, dyne/cm.^{2}	–	9.75×10^8	1.05×10^9
Shearing strength, dyne/cm.^{2}, parallel	–	4.9×10^8	–
perpendicular	–	1.16×10^9	5.55×10^8

*Lamé elastic moduli

described for most purposes as a lumped parameter system. Resonances are observed which can be attributed to the interaction of tissue masses with purely elastic structures. For higher frequencies, through the audio-frequency range and up to about 100 kc.p.s., the wave propagation of vibratory energy becomes more and more important but the type of wave propagation (shear waves, surface waves, or compressional waves) is strongly influenced by boundaries and geometrical configurations. Above 100 kc.p.s. and up into the mc.p.s. range, compression waves predominate and are propagated in a beam-like manner. This viewpoint permits not only a phenomenological description of the body's mechanical properties but, in an increasing number of cases, forms the basis for attempts to explain the behavior of tissue in terms of microscopic tissue- and cell-structure.

The Low Frequency Range. - Simple mechanical circuits such as one shown in figure 7 for a standing man, are usually sufficient to describe and understand the important features of the response of the human body to low frequency vibrations (34,36,37). Nevertheless it is difficult to assign numerical values to the elements of the circuit, since they depend critically on the kind of excitation, the body type of the subject, his position and muscle tone.

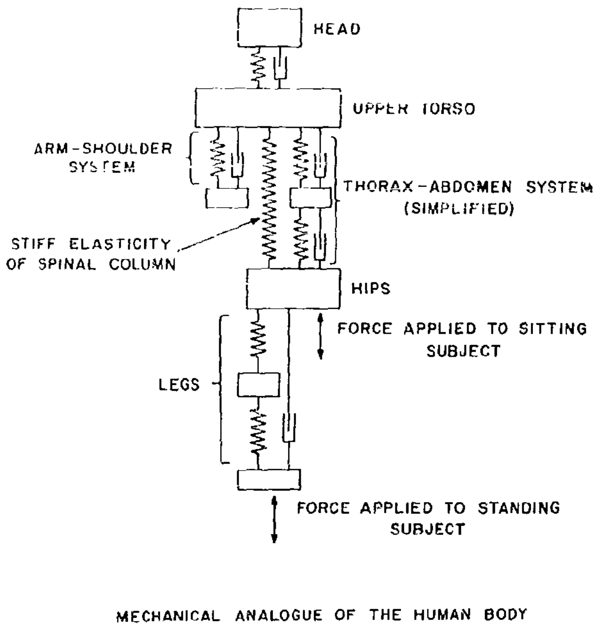


Figure 7. - Simplified mechanical circuit for a man standing on a vertically vibrating platform [after Goldman (34)].

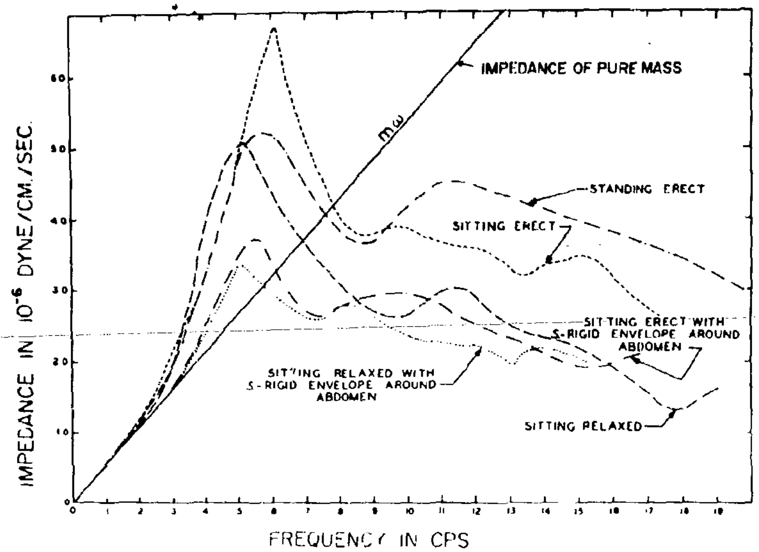


Figure 8. - Mechanical impedance of standing and sitting human subject vibrating in the direction of his longitudinal axis. The effects of body posture and of a semi-rigid envelope around the abdomen are also shown [after Coermann (32) and Coermann *et al.* (33)].

Subject exposed to vibrations in the longitudinal direction. - The mechanical impedance of a man standing or sitting on a vertically vibrating platform is shown in figure 8. Below approximately 2 c.p.s. the body acts as a unit mass. For the sitting man the first resonance is found between 4 and 6 c.p.s.; for the standing man resonance peaks are observed at about 5 and 12 c.p.s. (7,32). The numerical value of the impedance along with its phase allows calculation of the total energy transmitted to the subject.

The resonances at 4 to 6 c.p.s. and 10 to 14 c.p.s. are suggestive of mass-spring combinations of the entire torso with the lower spine and pelvis on the one hand and the upper torso on the other hand with forward flexion movements of the upper vertebral column. The assumption that flexion of the upper vertebral column occurs is supported by observations of the transient response of the body to vertical impact loads and associated compression fractures. The greatest loads occur in the region of the eleventh thoracic to the second lumbar vertebra which can therefore be assumed as the hinge area for flexion of the upper torso. Since the center of gravity of the upper torso is considerably forward of the spine, flexion movement will occur even with the force applied parallel to the axis of the spine (see also fig. 23). Changing the direction of the force so that it includes an angle with the spine (for example by tilting the torso forward) influences this effect considerably. Similarly the center of gravity of the head can be considerably in front of the neck joint which permits forward-backward motion. This situation results in forward-backward rotation of the head instead of pure vertical motion. Examples of relative amplitudes for different parts of the body are shown in figure 9 for the

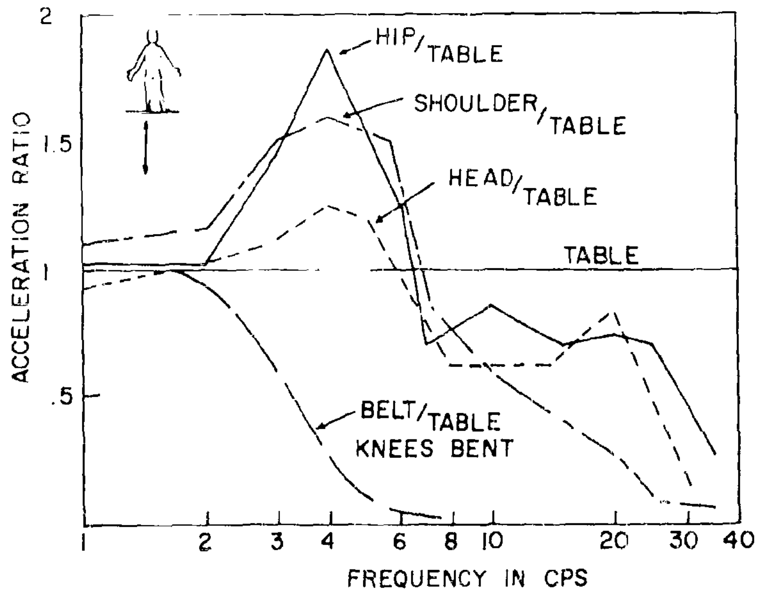


Figure 9. - Transmission of vertical vibration from table to various parts of the body of a standing human subject [after Dieckmann (7), data for transmission to belt after Radke (35)].

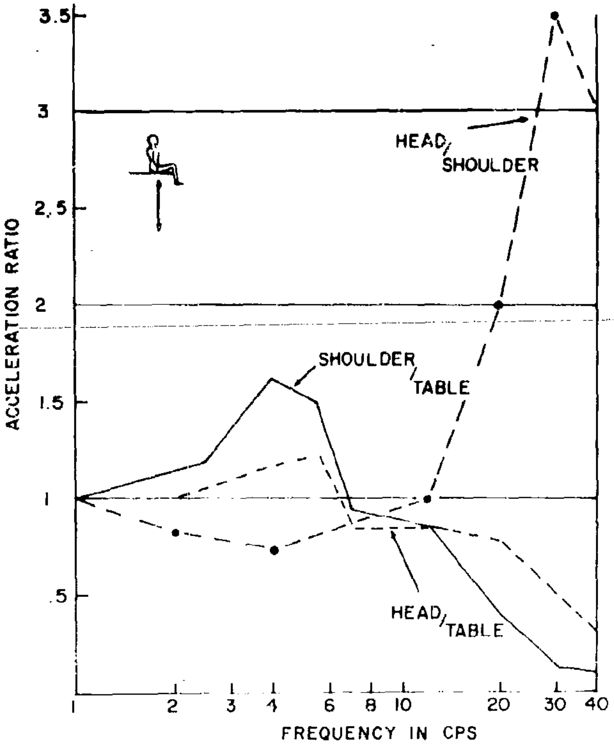


Figure 10. - Transmission of longitudinal vertical vibration from table to various parts of body of seated human subject [after Dieckmann (7)].

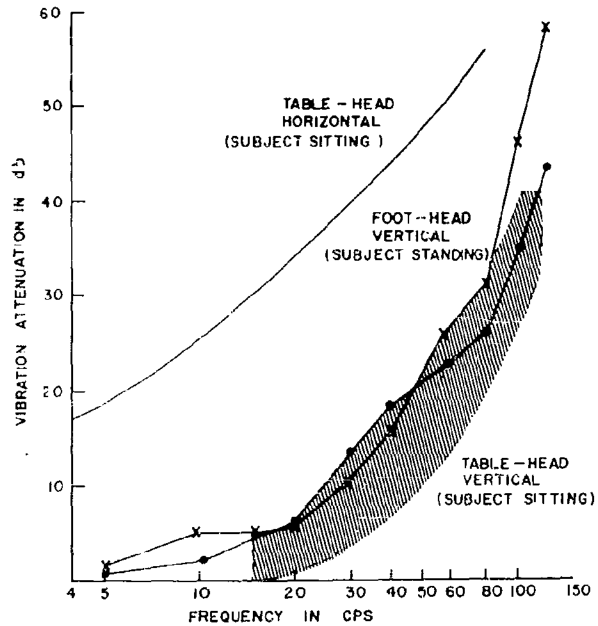


Figure 11. - Attenuation of vertical and horizontal vibration for standing and sitting human subjects. [Continuous lines after von Békésy (2)]. Shaded area is range of values for 10 subjects [after Coermann (36)].

standing and in figure 10 for the sitting subject (7). The curves show an amplification of motion in the impedance resonance range and a decrease at higher frequencies. The impedances and the transmission factors are changed considerably by individual differences in the body and its posture as well as support by a seat or back rest for a sitting subject or by the state of the knee or ankle joints of a standing subject. The resonance frequencies remain relatively constant whereas the transmission ratio varies [for the condition of fig. 10 transmission factors as high as 4 have been observed at 4 c.p.s. (35)]. Above approximately 10 c.p.s. vibration amplitudes of the body are smaller than the amplitudes of the exciting table and decrease continuously with increasing frequency. The attenuation of the vibrations transmitted from the table to the head is illustrated in figure 11. At 100 c.p.s. this attenuation is around 40 db. The attenuation along the body at 50 c.p.s. is shown, although not for pure longitudinal excitation, in figure 12.

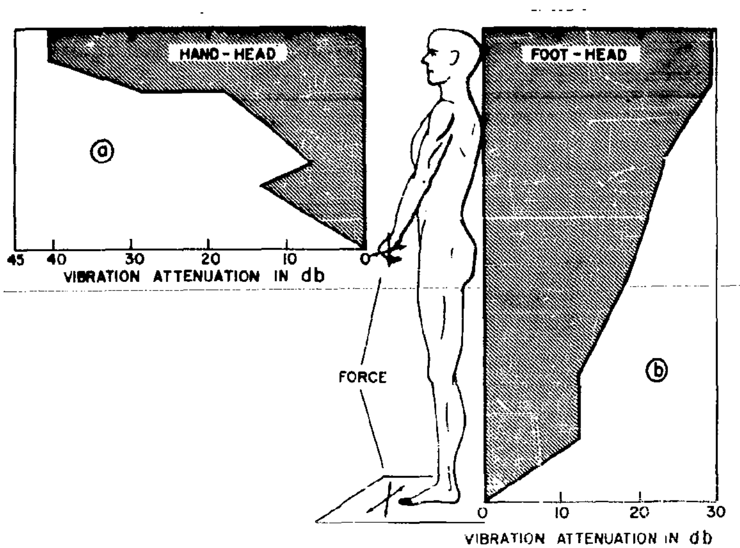


Figure 12. - Attenuation of vibration at 50 c.p.s. along human body. Excitation of a) hand and b) platform on which subject stands [after von Békésy (2)].

Between 20 and 30 c.p.s. the head exhibits a resonance as can be seen clearly in figures 9 and 10. In this range the head amplitude can exceed the shoulder amplitude by a factor of 3. This resonance is of importance in connection with the deterioration of visual acuity under the influence of vibration. Another frequency range of disturbances between 60 and 90 c.p.s. suggests an eyeball resonance (36).

The impedance of the human body lying on its back on a rigid surface and vibrating in the direction of its longitudinal axis has been determined by ballistocardiograph studies (37). The total mass of the body forms a simple mass-spring system with the elasticity and resistance of the skin for tangential vibration. For the average subject the resonant frequency is between 3 and 3.5 c.p.s. and the Q of the system is about 3. Restricting the subject's mobility by clamping the body at the feet and shoulders between plates connected with the table changes the resonant frequency to approximately 9 c.p.s. and the Q to about 2.5.

One of the most important subsystems of the body, which is excited in the standing and sitting position as well as in the lying position is the thorax-abdomen system (37,38). The abdominal viscera have a high mobility due to the very low stiffness of the diaphragm and the air volume of the lungs and the chest wall behind it. Under the influence of both longitudinal and transverse vibration of the torso, the abdominal mass vibrates in and out of the thoracic cage. These vibrations not only take place in the (longitudinal) direction of excitation but during that phase of the cycle when the abdominal contents swing towards the hips the abdominal wall is stretched outward and the abdomen appears larger in volume; at the same time the downward deflection of the diaphragm causes a decrease of the chest circumference. At the other end of the cycle the abdominal wall is pressed inward, the diaphragm upward and the chest wall is expanded. This periodic displacement of the abdominal viscera has a sharp resonance between 3 and 3.5 c.p.s. (fig. 13). It should be clear that the oscillations of the abdominal mass are coupled with the air oscillations of the mouth-chest system (19,33). Measurements of the impedance of the latter system at the mouth by applying oscillating air pressure to the mouth shows that the abdominal wall and the anterior chest wall respond to this pressure. The impedance has a minimum and the phase angle is zero between 7 and 8 c.p.s. The abdominal wall shows maximum response between 5 and 8 c.p.s., the anterior chest wall between 7 and 11 c.p.s. Vibration of the abdominal system resulting from exposure of a sitting or standing subject is clearly detected as modulation of the flow velocity through the mouth (fig. 13). (At large amplitudes speech can be modulated at the exposure frequency.) An electrical equivalent circuit for the abdomen-chest-mouth system is shown in figure 14.

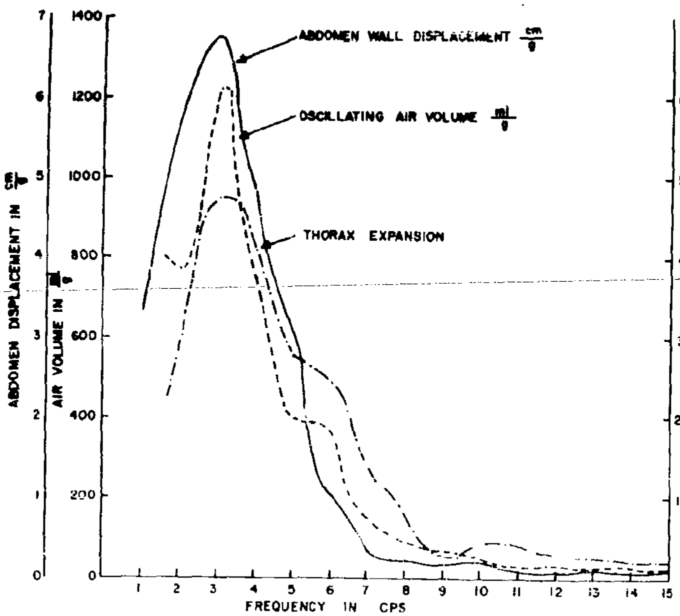
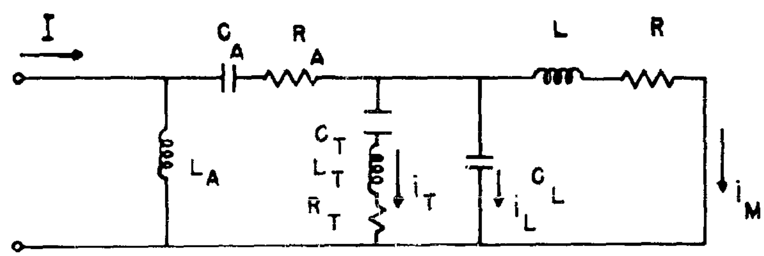


Figure 13. - Typical response curves of the thorax-abdomen system of a human subject in the supine position exposed to longitudinal vibrations. The displacement of the abdominal wall (2 in. below umbilicus), the air volume oscillating through the mouth, and the variations in thorax circumference are shown per g longitudinal acceleration [from Coermann *et al.* (33)].

Figure 14. - Electrical analog circuit for abdomen-chest-mouth system. L_A is abdominal mass, C_A is elasticity of abdominal wall, diaphragm, etc., R_A is resistance associated with abdominal mass; C_T, L_T, R_T are elasticity, mass, resistance of thorax; C_L is elasticity of air volume of lungs; L, R are mass and resistance for air flow into and out of the lungs. I is excitation velocity of torso; i_T, i_L, i_M are velocities of wall of thorax, compression of lung and of air flow.



Subject exposed to vibrations in the transverse direction. - The physical response to transverse vibration –i.e., horizontal in the normal upright position–is quite different from that described for vertical vibration: instead of thrust forces acting primarily along the line of action of the force of gravity on the human body, they act at right angles to this line. The distribution of the body masses along this line is therefore of the utmost importance and greater differences must be expected between sitting and standing subjects than for vertical vibration where the supporting structure of the skeleton and especially the spine have been designed for vertical loading.

Impedance measurements for transverse vibration are not available. The transmission of vibration along the body is illustrated in figure 15 (39). For a standing subject the hip, shoulder, and head amplitudes are of the order of 20 to 30 percent of the table amplitude at about 1 c.p.s. and decrease with increasing frequency. Relative maxima of shoulder and head amplitudes occur at 2 and 3 c.p.s. respectively. The sitting subject exhibits amplification of the hip (1.5 c.p.s.) and head (2 c.p.s.) amplitudes. All critical resonant frequencies appear to be between 1 and 3 c.p.s. Investigation of experimental results of the type of figure 15 in connection with phase measurements shows that the transverse vibration patterns of the body can be described as standing waves, i.e., as a rough approximation one can compare the body with a rod in which transverse flexural waves have been excited. One has, therefore, in agreement with the experimental results, nodal points on the body which become closer to the feet as the frequency increases since the phase shift between all body parts and the table increases continuously with increasing frequency. At the first characteristic frequency at 1.5 c.p.s. the head of the standing subject is observed to have a 180° phase shift compared with the table; between 2 and 3 c.p.s. this phase shift became 360° (39).

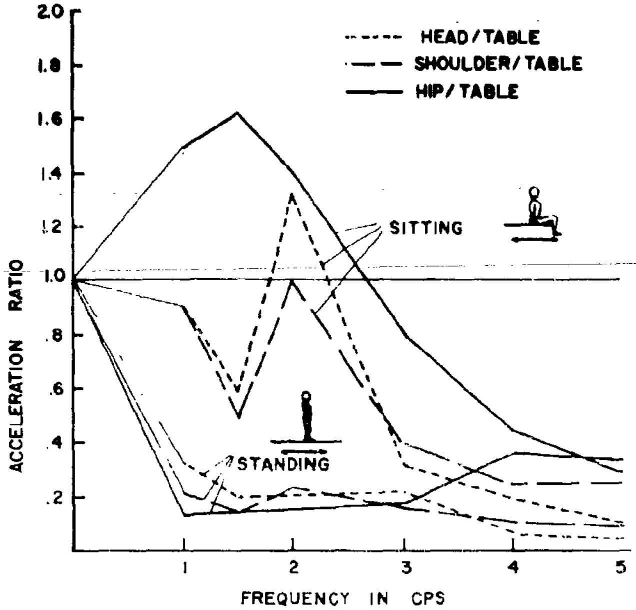


Figure 15. - Transmission of transverse, horizontal vibration from table to various parts of sitting and standing human subject [after Dieckmann (39)].

Note that there are longitudinal head motions excited by the transverse vibration in addition to the transverse head motions shown in figure 15 and discussed above. The head performs nodding motion due to the anatomy of the upper vertebrae and the location of the head's center of gravity. Above 5 c.p.s. the head motion for the sitting and standing subjects is predominantly vertical (of the order of 10 to 30 percent of the horizontal table motion).

Vibrations transmitted to the hand. - In connection with studies on the use of vibrating hand tools several measurements have been made of vibration transmission from hand to arm and body (20,40,41). The impedance measured on a hand grip for a specific condition representative of hand tool use is presented in figure 16. The impedance has one maximum in the range below 5 c.p.s., probably determined by the

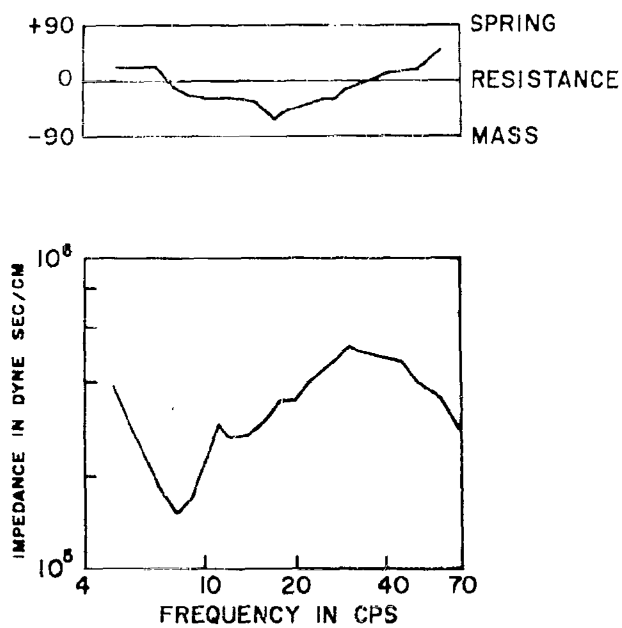


Figure 16. - Impedance and phase angle of arm measured at a vibrating hand grip. Elbow flexion 20° to 25°, static pressure on grip 22 lbs. Measurements on one subject [after Dieckmann (40)].

natural frequencies for transverse excitation of the human body between 1 and 3 c.p.s. A second strong maximum appears between 30 and 40 c.p.s.; the effective mass of the hand [approximately 2.2 lbs. (1 kg.)] is here in resonance with the elasticity of the soft parts on the inside of the hand. This elasticity between force and hand has been estimated as 2.10^{-8} cm/dyne. With a practical hand tool, which operates between 40 and 50 c.p.s., it has been found that the vibration amplitude decreases from the palm to the back of the hand by 35 to 65 percent. Further losses occur between the hand and the elbow and the elbow and the shoulder. Figure 16 which shows this decrease of vibration amplitude from the hand to the head, indicates that the strongest attenuation occurs in the shoulder joint.

The Middle Frequency Range (*Wave Propagation*). - Above roughly 100 c.p.s. simple lumped parameter models become more and more unsatisfactory for describing vibrations of tissue and it becomes necessary to look at the tissue as a continuous medium for sound propagation.

Skull vibrations. - The vibration pattern of the skull agrees approximately withthe pattern of a spherical elastic shell (42,43). The nodal lines observed suggest that the fundamental frequency lies between 300 and 400 c.p.s. with resonances forthe higher modes around 600 and 900 c.p.s. The frequency ratio between the modes is approximately 1.7 while the theoretical ratio for a sphere is 1.5. From the observed resonances the elasticity of skull bone can be calculated. The value obtained for Young's modulus (1.4×10^{10} dynes/cm.^{2}) agrees fairly well with static test results on dry skull preparations, but is somewhat lower than the static test data obtained on the femur (table 2). Impedances of small areas on the skull over the mastoid area (44) have been measured for practical problems (table 1). The impedance of the skin lining in the auditory canal has been investigated and used in connection with studies on ear protectors (45).

Vibrations of the lower jaw with respect to the skull can be explained by a simple mass-spring system, which has a resonance, relative to the skull, between 100 and 200 c.p.s. (46).

Impedance of soft human tissue. - Impedance measurements of small areas (1 to 17 cm.^{2}) over soft human body tissue have been made with vibrating pistons between 10 and 20 kc.p.s. This impedance starts out at low frequencies as a large elastic reactance. With increasing frequency the reactance decreases, becomes zero at a resonance frequency and becomes a mass reactance with still further increase in frequency (fig. 17) (3,4,47). These data cannot be explained by simple lumped parameter models, but require a theory of wave propagation in a visco-elastic medium such as the tissue constitutes for this frequency range (47,48). The high viscosity of the medium makes possible the use of simplified theoretical assumptions, such as a homogeneous isotropic infinite medium and a vibrating sphere instead of a circular piston. The results of

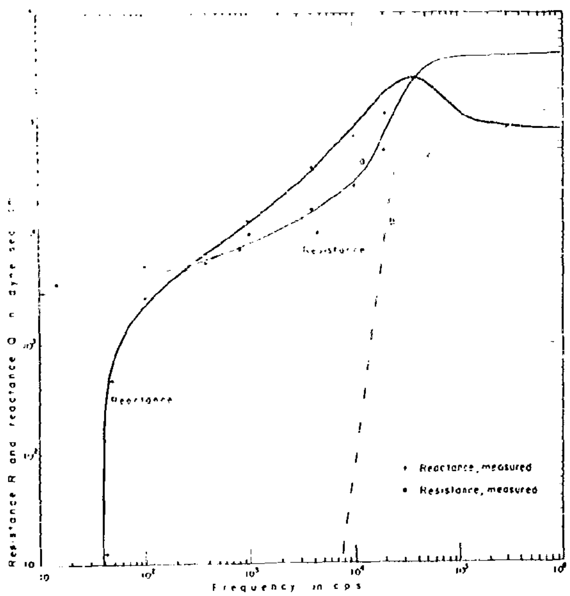


Figure 17. - Resistance and reactance of circular area, 2 cm. diameter, of soft tissue body surface. Smooth curves calculated for 2 cm. sphere vibrating in a) visco-elastic medium with properties similar to soft tissue (parameters as in table 2), b) frictionless compressible fluid, c) incompressible viscous fluid [from von Gierke *et al.* (47)]

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such a theory agree well with the measured characteristics. As a consequence it has been possible to assign absolute values to the shear viscosity and the shear elasticity of soft tissue (table 2). The theory together with the measurements shows that over the audio-frequency range, most of the vibratory energy is propagated through the tissue in the form of transverse shear waves and not in the form of longitudinal compression waves. Such shear waves have a much smaller propagation velocity (and therefore wave length) than sound and a strong dispersion. The velocity is about 20 m/sec. at 200 c.p.s. and increases approximately with the square root of the frequency. This may be compared with the constant sound velocity of about 1500 m/sec. for compressional waves. Some energy is propagated along the body surface in the form of surface waves which have been observed optically. Their velocity is of the same order as the velocity of shear waves (47).

From the mechanical impedance of the body surface one can calculate the acoustic absorption coefficient. This indicates what percentage of an incident air-borne sound wave is absorbed at the body surface and propagated through the tissue and what percentage is reflected (4). At 100 c.p.s. a small area of the forehead or of soft tissue absorbs only about 2 percent of the incident sound energy. At higher frequencies a still smaller percentage is absorbed. Only the specialized structure of the ear allows a small area of the body surface, the tympanic membrane, to absorb much more energy, for example, at 1000 c.p.s.: 50 to 80 percent. This is achieved by the middle ear transformer action, which matches the tissue structures of the inner ear to the characteristic impedance of air.

Ultrasonic Vibrations. - Above several hundred kc.p.s. in the ultrasound range, most of the vibratory energy is propagated through tissue in the form of compressional waves and geometrical acoustics offers a good approximation for the description of their path. Since the tissue dimensions under consideration are almost always large compared with the wavelength (about 1.5 mm. at 1 mc.p.s.) the mechanical impedance of the tissue is equal to the characteristic impedance, i.e., sound velocity times density. This value for soft tissue differs only slightly from the characteristic impedance of water (49). The most important factor in this frequency range is the tissue viscosity, which brings about an increasing energy absorption with increasing frequency (48). At very high frequencies this viscosity also generates shear waves at the boundaries of the medium, at the boundary of the acoustic beams, and in the areas of wave transition to media with somewhat different constants, (e.g., boundary muscle to fat tissue, or soft tissue to bone). These shear waves are attenuated so rapidly that they are of no importance for energy transport but are noticeable as increased local absorption, i.e., heating.

From 500 kc.p.s. to 10 mc.p.s. the attenuation coefficient describing the decrease of the sound intensity in a plane ultrasound wave, is only in fair agreement with the value one would calculate from the tissue viscosity measured in the audio-frequency range (table 2). The tissue deviates in this frequency range from the behavior of a medium with constant viscosity. In figure 18 attenuation coefficients measured in different types of tissue are summarized (49,50). (On this graph a slope $\frac{\alpha}{f^2}$ = constant (where f is the frequency) would be indicative of classical viscous absorption with constant shear viscosity. A smaller slope, or a change in slope, indicates a change in viscosity with frequency (relaxation phenomenon). The graph gives only a few examples and typical functions from a large body of attenuation data available. The absorption of most soft tissues is in the range from 0.5 to 2 db/cm/mc.p.s. The order of increasing absorption is: brain tissue, liver tissue, striated muscle, smooth muscle, kidney, skin and tendon. Bone has the highest value with approximately 10 db/cm. The ultrasonic absorption coefficient depends very much on the structural features of the tissue and might well aid in obtaining a clearer quantitative picture of the mechanical structure of cells. Interesting in this respect are the acoustic anistropies, i.e., cases where attenuation depends on the direction of propagation: fiber anistropy has been found in the collinear fibers of striated muscle (fig. 18) and layer anistropy can occur in structures consisting of parallel layers of different tissue types such as in the abdominal wall.

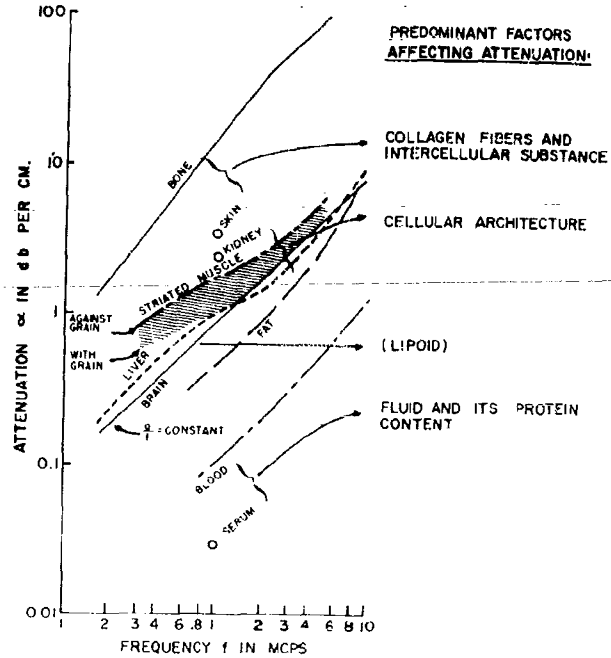

Figure 18. - Approximate values of high frequency sound attenuation in various tissues [after Goldman and Hueter (49), Dussik *et al.* (50)].

MECHANICAL DATA FROM SHOCK FORCES

At the present time, very little in the way of numerical data on mechanical characteristics is available from studies on shock or impact forces. Some evidence of resonances has been noted (51) but much of this is better obtained from vibration studies. The application of mechanical data which has been obtained from studies on vibration to exposure to shock and impact will be discussed in the section on mechanical damage. Mechanical responses to shock and impact are, in general, extremely difficult to analyze numerically for basic body characteristics.

EFFECTS OF SHOCK AND VIBRATION

The motions and stresses resulting from the application of mechanical forces to the human body have several possible effects: 1) the motion may interfere directly with physical activity; 2) there may be mechanical damage or destruction; and 3) there may be secondary effects which operate through biological receptors and transfer mechanisms and produce changes in the organism including subjective phenomena. Thermal and chemical effects are uncommon.

Mechanical Interference. - The relationships between applied forces and the motion of the body and its parts have been discussed above. There are many ways indeed in which the forces can be applied and the body itself can take on many attitudes. Certain types of displacement, velocity or acceleration, if of sufficient magnitude, can be very disturbing to sensory and neuromuscular activities such as reading instruments or making fine adjustments of controls or of the position of the body and its parts. In particular, speech communication may be rendered difficult. Very little is known about how much of what kind of motion interferes with particular activities. Further, such information, when available, has meaning only in terms of tolerances permitted for the activities in question. For example, the disturbance of visual acuity arising from body vibration is not only frequency dependent (36) (fig. 19) but may be expected to be roughly proportional to the amplitude of the vibration and may be dealt with by changing the frequency, reducing the amplitude or by decreasing the acuity required for the given task. Mechanical motion resulting from occasional shocks will not be expected to interfere directly with most tasks unless timed critically with respect to some operation or unless repeated at very short intervals. Such interference with task performance as does arise from shocks is more apt to be the result of biological responses or of actual damage.

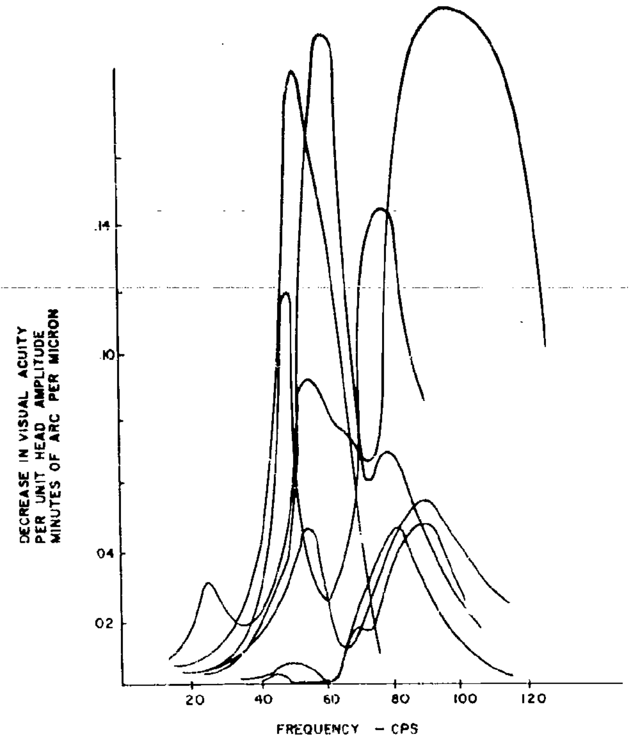

Figure 19. - Decrease of visual acuity of observers standing on vibrating platform [after Coermann (36)]. Each curve represents a different subject.

Mechanical Damage. - Of the many kinds and degrees of mechanical damage to the human body arising from the application of mechanical forces, certain kinds have been singled out for attention and study because they are particularly common, dangerous or disturbing in some special way. Among these, short of actual destruction, are bone fracture, lung damage, injury to the inner wall of the intestine, brain injury, cardiac damage, ear damage, tearing or crushing of soft tissues, and certain special types of chronic injury such as tendon or joint strains and the "white finger" syndrome of vibrating tool operators. These will be considered in detail later on.

Biological Responses. - Mechanical stresses and motions may stimulate various receptor organs in the skin and elsewhere or may excite parts of the nervous system directly. The result may be reflex activity or modification of it. The stimuli initiate nervous system and hormonal activity which has a marked modifying action on many metabolic processes relating to food assimilation, muscular activity, reproductive activity, etc. (52). These changes are difficult to measure and correlate and seem to differ in different species, at least in degree. Nevertheless, considerable indirect evidence exists for the reality of these response patterns and it is generally agreed that the broad picture arrived at from animal experimentation applies to man. Exposure to mechanical forces, when of great enough extent and duration, must certainly assume part of the responsibility for phenomena such as fatigue, changes in work capacity, ability to maintain attentiveness, etc. In response to acute stimulation, excitation of brain centers may produce emotional reactions such as fear or unpleasantness and lead to automatic or deliberate compensatory or protective behavior.

EFFECTS OF MECHANICAL VIBRATION

Mechanical Damage is produced when the accelerative forces are high enough. However, it is obvious that experimental data can only be obtained from work on animals and that the results must be handled very carefully in extending them to humans. Mice (53), rats (54), and cats (55,56) have been killed by exposure to vibration. There is a definite frequency dependence of the lethal accelerations which coincides with the resonant displacement of the visceral organs, but which has been only partly established. Mice are killed

at 10 to 20g within a few minutes in the range 15 to 25 c.p.s.; above and below this frequency range, the survival time is longer. Rats and cats are likewise killed within 5 to 30 minutes at accelerations above about 10g but the frequency dependence has not been worked out. Post-mortem examination of these animals usually shows lung damage, often heart damage, and occasionally brain injury. The injuries to heart and lungs probably result from the beating of these organs against each other and against the rib cage. The brain injury, which is a superficial hemorrhage, is not yet interpretable in definite terms; it may be due to relative motion of the brain within the skull, to mechanical action involving the blood vessels or sinuses directly or to secondary mechanical effects. Tearing of intra-abdominal membranes is rarely seen. Exposure for several minutes to peak acceleration of about 5g often produces heart damage as indicated by delayed changes in the electrocardiogram (56). An increase in body temperature is found on exposure to vibration. Since this occurs also in dead animals it is probably mechanical in origin. Calculations of heat absorption based on body impedance data suggest that appreciable heat can be generated at large amplitudes. Exposure of monkeys to 5g at 10 and 20 c.p.s. for several hours seems to produce some damage to the vestibular system but these findings require confirmation (57). Observations on man (58) have been made in a few instances and indicate that above about 3g, sharp pain in the chest may occur. Traces of blood have occasionally been found in the feces after exposure to 6g at 20 to 25 c.p.s. for about 15 minutes. This suggests mechanical damage to the intestine or rectum.

It is clear that several of the phenomena found in animals must also be possible in humans. Mechanical damage to heart and lungs, injury to the brain, tearing of membranes in the abdominal and chest cavities, as well as the above mentioned intestinal injury—all these seem possible. So also does the heating of the body when shaken. Unfortunately acceleration-frequency curves for these effects have not been established. Because of the relatively greater visceral masses of the human, the minima of such curves, which would correspond to resonance ranges, must occur at relatively lower frequencies than in small animals. As to whether the amplitudes required may be larger or smaller than in animals, it is not yet possible to say. Subjective symptoms such as the occurrence of chest pain after exposure to 3g (table 3) may or may not be significant although from the point of view of safety they must be taken seriously until more is known about the details of the processes involved.

Table 3. Criteria of Tolerance for Short Exposure to Vertical Vibration
[From Ziegenruecker and Magid (74), see figure 21]
Each Cross Indicates a Decided Comment from a Human Subject as to His Experiencing the Symptom Listed.

Symptom c.p.s.	A	B	C	D	E	F	G
1					xxxxx xxx		xxx
2					xxxxx xxx		xxxx
3	xx	xx			xxxxx	x	xxxxx
4	xx	xx		xx	xxx	xx	xxxxx
5		xxxx				x	xxxxx x
6	xxx	xxxx		x			xxxx
7	xx	xxxxx	x	x			x
8	x	xxxx		x		xx	xxx
9	xx	xxxx			x		xxxxx
10	x	x	xxx	xx		x	
15							xxxxx xxx

A: Abdominal Pain
B: Chest Pain
C: Testicular Pain
D: Head Symptoms
E: Dyspnea
F: Anxiety
G: General Discomfort

Chronic injuries may be produced by vibration exposure of long duration at levels which produce no apparent acute effects. In practice such effects are usually found after exposure to repeated blows or to random jolts rather than to sinusoidal motion. When such shocks or blows are applied to the human body at relatively short intervals, the relation of the interval to tissue response times becomes very important. Exposure to such forces frequently occurs in connection with the riding of vehicles. Buffeting in aircraft or in high speed small craft on the water, and shaking in heavy vehicles on rough surfaces, give rise to irregular jolting motion. Acute injuries from exposure to these situations are rare but complaints of discomfort and chronic minor injury are common. Truck and tractor drivers often have sacro-iliac strain. Minor kidney injuries are occasionally suspected and, rarely, traces of blood may appear in the urine. The length of exposure and the details of the ways in which the body is supported play an important role.

Chronic injuries are also produced by localized vibration. A classical example of this is the pain and numbing of the fingers on exposure to cold which affects many people after several months of using such equipment as pneumatic hammers and drills or hand-held grinders or polishers. The heavier, slow moving devices appear to produce more severe jolting. There is an extensive clinical literature on this subject (59) but little known of the mechanism of the injury or of the actual forces responsible although many high frequency components may be present (41). The repeated insults to the tissues seem gradually to affect the capillaries and their nerve supply. Injuries resembling this have been produced in the feet of rats exposed to 60 c.p.s. at 8 to 9g for 10 to 12 hours per day up to about 1000 hours (60).

Physiological Responses have been studied very little. Rats exposed for 10 to 40 minutes a day for several months to about 15g at 12.5 c.p.s. appear to show minor behavior abnormalities (61). The adrenal glands of rats show a rapid fall in ascorbic acid content on exposure to accelerative levels of a few tenths of a *g* unit at 5 to 10 c.p.s. (62), and changes in the reproductive cycle and growth have been observed in rabbits exposed to vibration for several days (64). Changes in respiration, heart activity and peripheral circulation have been observed in both men and animals as immediate and possibly transient responses to moderate vibration. These results indicate the need for further study. Certain postural reflexes appear to be inhibited by vibratory motion (64,65).

Subjective Responses to vibration include perception, feelings of discomfort, apprehension and pain. Since acceleration, frequency, mode of application, duration and the situation of the subject are all involved, it is exceedingly difficult to find simple definitive ways of characterizing results. Early workers in this field (66-71) concerned themselves with whole-body exposure under conditions which were believed to be of practical interest. The results which they obtained were rough and ready and there appears to have been little control of subjective factors. Generally there are three apparently simple criteria: the thresholds of perception, of unpleasantness, and of tolerance. The two latter are difficult to identify and reproduce although agreement to within a factor of about 3 has been obtained. A compilation of these results based on exposures of about 5 to 20 minutes is given in figure 20 (72). For longer exposures, data are limited. Some information has been obtained on comfort and tolerance levels for aircraft pilots (73). Very long exposure to vibration much above the level of perception seems to be irritating and fatiguing.

For short exposures, less than 5 minutes, a study has been carried out in the frequency range 1 to 15 c.p.s. (74). Subjects were exposed to a specified acceleration until they could no longer tolerate it. They were then asked for their reactions and what their specific reason was for asking to be released. Table 3 shows a distribution of the major reasons for each of the frequencies used. Evidently no single criterion of tolerance was found although some reactions were more common than others. The estimated limits of tolerance for short exposures according to these criteria are given in figure 21.

An interpretation of feelings of discomfort and apprehension arising from exposure to vibration can be based on the idea that vibration of certain organs produces nervous system excitation directly responsible for these feelings. If one then were to apply to the body a vibratory excitation of a specified frequency and amplitude, one could knowing the mechanical characteristics of the body and its parts, estimate the response of the organs of interest and thus establish a basis for rational tolerance limits. Where large visceral masses are involved, there is already some information available from which such calculations can be made. However, the detailed application of this principle requires more data than are now at hand.

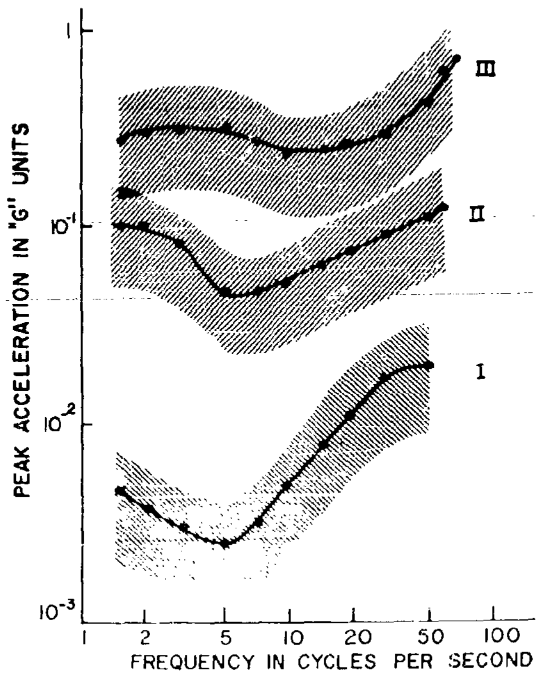

Figure 20. - Average peak accelerations at various frequencies at which subjects perceive vibration (I), find it unpleasant (II), or refuse to tolerate it further (III). Exposures of 5 to 20 minutes. Shaded areas are about one standard deviation on either side of mean. Data averaged from seven sources [from Goldman (72)].

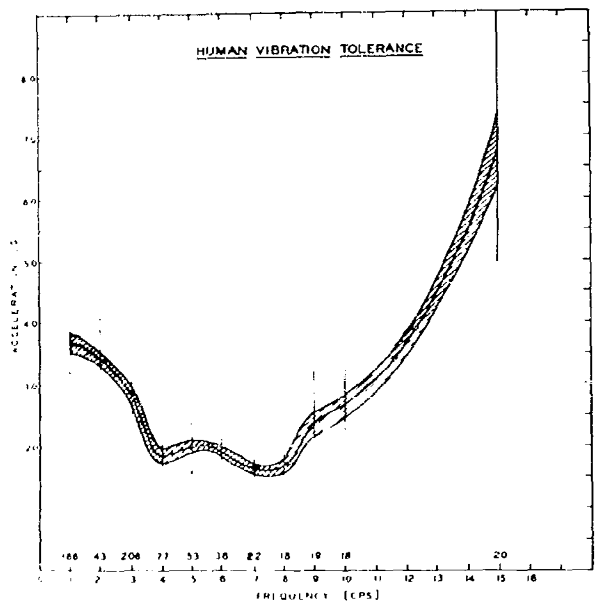

Figure 21- Average peak acceleration at various frequencies at which subjects refuse to tolerate further a short exposure to vertical vibration. The figures above the abscissa indicate the exposure time in seconds at the corresponding frequency. The shaded area has a width of one standard deviation on either side of the mean (10 subjects) [from Ziegenruecker and Magid (74)].

EFFECTS OF MECHANICAL SHOCK

Mechanical shock includes several types of force application which have similar, though not identical, effects. Explosions, explosive compression or decompression, impacts and blows from rapid changes in body velocity or from moving objects produce shock forces of importance. Major damage, short of complete tissue destruction, is usually to lungs, intestines, heart or brain. Differences in injury patterns arise from differences in rates of loading, peak force, duration, localization of forces, etc.

Blast and Shock Waves (5,17,75,110). The mechanical effects associated with rapid changes in environmental pressure are primarily localized to the vicinity of air-filled cavities in the body, i.e., the lungs and the air-containing gastro-intestinal tract. Heavy masses of blood or tissue border here on light masses of air and the local impedance mismatch can lead to destructive relative tissue displacement by several different mechanisms. Starting with very slow differential pressure changes, of approximately 1-second duration or longer, dynamic mechanical effects are unimportant; the static pressure is responsible for destructive mechanical stress or physiological response. Such pressure time functions occur with the explosive decompression of pressurized aircraft cabins at high altitude and with the slow response of well sealed

shelters to blast waves. If the pressure rise or fall times are shortened (roughly to the order of tenths of seconds), the dynamic response of the different resonating systems of the body becomes important, in particular the thorax-abdomen system of figure 14, but now with the force applied as in figure 22. The dynamic

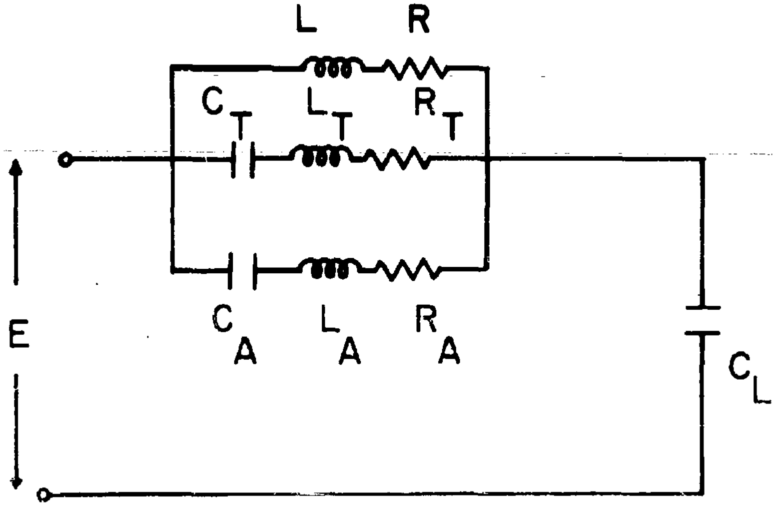

Figure 22. - Electrical analogue circuit of abdomen-chest-mouth system of figure 14, but with excitation (E) by environmental pressure changes. (Lettering as in fig. 14.)

load factor of the specific blast disturbance under consideration determines the response. The meager data on the effect of known blast loads on man and animals do not permit evaluation and recognition of the different responsive systems and the biologically most effective pressure time curve, i.e., the one with the minimum peak pressure. Available data for single pulse, instantaneously rising pressures point towards the existence of such a minimum which corresponds to natural frequencies for dogs of between 10 and 25 c.p.s. (17); for humans this frequency would be lower. For pressures with total durations of milliseconds or less and much shorter rise times (duration of wave short compared with the natural period of the responding tissue) the effect and destruction seems to depend primarily on the momentum of the shock wave. The mass m of an oscillatory system located in a wall or body surface struck by a shock wave will be set into motion according to the relation: $\int P_r dt = mv_0$ with P_r = reflected pressure at body surface, t = time and v_0 = initial velocity. Experimental fatality curves on animals (5) generally show this dependence on momentum for short pressure phenomena (close to center of detonation) and the transition to a dependence on peak pressure for phenomena of long duration (far away from center). Fatal blast waves in air and water, for example, differ widely in peak pressure and duration (in air 10 atm. in excess of atmospheric pressure with a duration of 2.8 millisec., in water 135 atm. in excess with a duration of 0.17 millisec.) but their momenta are similar. In this most important range of short duration blasts the mechanical effects are localized due to the short duration, i.e., the high frequency content of the wave. The upper respiratory tract and bronchial tree as well as the thorax and abdomen system are too large and have too low resonating frequencies to be excited; there is no general compression or overexpansion of the thorax, which leads to pulmonary injury as in explosive decompression. The blast waves go directly through the thoracic wall producing an impact or grazing blow. Inside the tissue blast injury has three possible causes: 1) spalling effects, i.e., injuries caused by the tensile stresses arising from the reflection of the shock wave at the boundary between media with different propagation velocity (for example subpleural pulmonary hemorrhages along the ribs); 2) inertia effects which lead to different accelerations of adjacent tissues with various densities, when the shock wave passes simultaneously through these media; 3) implosion of gas bubbles enclosed in a liquid. These phenomena are similar to the observations made with high velocity missiles passing through water near air-containing tissues (76,77). The shock waves may produce not only pulmonary injuries but also hard, sharply circumscribed blows to the heart.

Of the injuries produced by exposure to high-explosive blast lung hemorrhage is one of the most common. It may not of itself be fatal, since enough functional lung tissue may easily remain to permit marginal gas

exchange. However, the rupture of the capillaries in the lung produces bleeding into the alveoli and tissue spaces which can seriously hamper respiratory activity or produce various respiratory and cardiac reflexes. The heart rate is often very slow after a blast injury. Leakage of fluid through moderately injured but not ruptured capillaries may occur. There is also the possibility that air may enter the circulation to form bubbles or emboli and by reaching critical regions may impair fatally the heart or brain circulation, or produce secondary damage to other organs (5). Fat emboli may also be formed and these, too, are capable of blocking vessels supplying vital parts. When gas pockets are present in the intestines, the shock may produce hemorrhage and in extreme cases rupture the intestinal wall itself.

The effects of underwater shock waves on man and animals are in general of the same kind as those produced by air blast. Differences which appear are those of magnitude and often depend on the mode of exposure of the body. A person in the water may, for example, be submerged from the waist down only, in which case damage is practically confined to the lower half of the body and intestinal, rather than lung damage, would occur (78).

Concussion damage to the heart itself is stated to be very rare but direct mechanical injury to the heart muscle and conducting mechanism is certainly possible. Cerebral concussion resulting directly from exposure to shock waves is unusual. Neurological symptoms following exposure to blast may however include general depression of nervous activity sometimes to the point of abolition of certain reflexes. Psychological changes such as memory disturbances and abnormal emotional states are sometimes found. In extreme cases, there may be paralysis or muscular dysfunction. Unconsciousness and subsequent amnesia for events immediately preceding the injury result more commonly from blows to the head than from air blast. Recovery from minor concussion may apparently be complete, but repeated concussion may produce lasting damage (79).

The ear is the part of the human body most sensitive to blast injury. Rupture of the tympanic membrane (17,80), and injury to the conduction apparatus can occur singly or together with injury to the hair cells in the inner ear (81). The two first mentioned injuries may protect the inner ear through energy dissipation. The degree of injury depends on the frequency content of the blast pressure function. The fact that the ear's greatest mechanical sensitivity is between 1500 and 3000 c.p.s. explains its vulnerability to short duration blast waves. Peak pressures of only a few pounds per square inch can rupture the ear drum and still smaller pressures can damage the conducting mechanism and the inner ear. There seem to be wide variations in individual susceptibility to these injuries.

Impacts, Blows, Rapid Deceleration. - This type of force is experienced in falls, in automobile or aircraft crashes, in parachute openings, in seat ejections for escape from high speed military aircraft and in many other situations such as certain sports. Interest in the body's responses to these forces centers on mechanical stress limits. The injuries which occur most often are bruises, tissue crushing, bone fracture, rupture of soft tissues and organs, and concussion. A bruise is a superficial area of slight tissue damage with rupture of the small blood vessels and accumulation of blood and fluid in and around the injured region. It is essentially a crushing injury produced by compression of the tissues, usually between the impinging solid and the underlying bone. It is extremely common and is readily repaired by the body itself. When the tissue is completely destroyed by crushing, the damage is usually irreparable. Bone fractures, like bruises, require that the forces be sustained for long enough to produce appreciable displacements and properly concentrated stresses.

When soft tissues are displaced considerable distances by appropriate forces, so-called internal injuries, i.e., rupture of membranes or organ capsules may take place. Such injuries are, in practice, more often produced by forces of relatively long duration and are usually dangerous.

Experiments have been carried out in which animals were embedded in plaster casts and then dropped about 20 ft. onto a metal plate which had various degrees of cushioning. In this case the decelerative impact was well distributed. The degree of injury increased with the acceleration. Lung hemorrhage and laceration of liver and spleen capsules were most common. Damage to the diaphragm, the brain, and the bone structure was seen when the deceleration exceeded about 500g (velocity change 35 ft/sec.) (82).

The obvious correlation between the response of the body system to continuous vibration and to spike and step-force functions has not until recently been used to guide and interpret experiments. The tissue areas stressed to maximum relative displacement at the various frequencies during steady state excitation are naturally preferred target areas for injury under impact load if the force-time functions of the impacts have appreciable energy in these frequency bands, i.e., if the impact duration is of the same order of magnitude as the body's natural periods. If the impact exposure times are shorter, stress tolerance limits increase; if exposure times decrease to hundredths or thousandths of a second, the response will become more and more limited and localized to the point of application of the force (blow). Elastic compression or injury will depend on the load distribution over the application area, i.e., the pressure to which tissues are subjected. If tissue destruction or bone fracture occur close to the area of application of the force these will absorb additional energy, protect deeper seated tissues by reducing the peak force and spreading it over a longer period of time. An example is the fracture of foot and ankle of men standing on the deck of warships when an explosion occurs beneath. The support may be thrown upward with great momentum and if the velocity reaches 5 to 10 cm/sec. (acceleration of several hundred g) fractures occur (83). However, the energy absorption by the fracture protects structures higher up.

If the force functions contain extremely high frequencies the compression effects spread from the area of force application throughout the body as compression waves. If these are of sufficient amplitude they may cause considerable tissue disruption. Such compression waves are observed from the impact of high velocity missiles.

If the exposure to the accelerating forces lasts long enough for the whole body to be displaced, exact measurement of the force applied to the body and of the direction and contact areas of application become of extreme importance. In studies of seat ejection, for example, knowledge of seat acceleration alone is not sufficient for estimating responses. One should know the forces in those structures or restraining harnesses through which the acceleration forces are transmitted. The location of the center of gravity of the various body parts like arms, head and upper torso must be known over the time of force application so that the resulting body motion and deformation can be explained and influenced for protection purposes. In addition to the primary displacements of body parts and organs there are secondary forces from decelerations if, due to the large amplitudes, the motions of parts of the body are stopped suddenly by hitting other body parts. Examples occur in linear deceleration where, depending on the restraint, the head may be thrown forward until it hits the chest or, if only a lap belt is used, the upper torso may jackknife and the chest may hit the knees. Under field conditions there is always the additional possibility that the body may strike nearby objects thus initiating a new impact deceleration history.

Longitudinal acceleration. - The study of positive longitudinal (headward) acceleration of short duration is closely connected with the development of upward ejection seats for escape from aircraft. Since the necessary ejection velocity of approximately 60 ft sec. (10 millisec.) and available distance for the catapult guide rails of about 3 ft. (1 millisec.) are determined by the aircraft without much leeway, the minimum acceleration required (step function) would be approximately 9g. Since the high jolt of the instantaneous acceleration increase is undesirable because of the high dynamic load factor of this function for the frequency range of the body resonances (compare with fig. 10), slower build-up of the acceleration with higher final acceleration has appeared preferable. Minimum dynamic overshooting is achieved with build-up times of about 0.1 sec. and total force times of approximately 0.2 sec. The predominant oscillation periods observed during such ejections are between 7 and 14 c.p.s. Recent investigations show that the body's ballistic response can be predicted with good accuracy by means of analogue computations making use of the body's continuous frequency response characteristic (84). Another approach has substituted, as a model for the seated body, a homogeneous elastic rod, one end of which is free. Assuming a wave travel time from the accelerated end to the free end of 0.025 sec. (resonant frequency of 10 c.p.s.) good agreement between experimental transient head acceleration and theoretical end accelerations of the rod has been obtained (85). Since the vertebrae between the eighth thoracic and fifth lumbar fracture at approximately the same static load, in the neighborhood of 25g, the tolerance limit of 20g presently generally assumed allows for minimum overshooting. Forceful flexion of the spine and neck must be controlled by either positioning or protective harnesses (fig. 23).

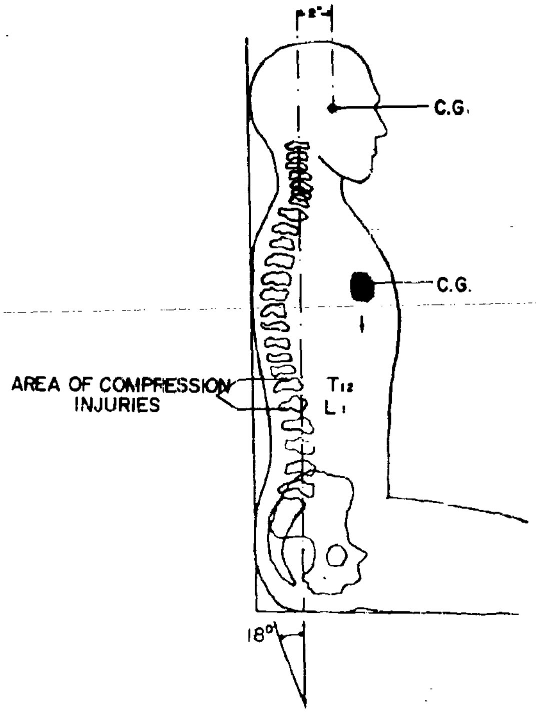

Figure 23. - Alignment of seated subject for positive, vertical acceleration or vibration. Center of gravity of head and of body above 12th thoracic vertebra are indicated.

In addition to forces which may bend the upper torso there is a turning moment which acts on the pelvis and increases the load on the spine since the contact point with the seat is not in line with the spine (86). If the tolerance limits are exceeded spinal fractures of the lumbar and thoracic vertebrae occur first: neck injuries apparently require considerably higher levels.

For vertical crash loads in general the same considerations apply as discussed for seat ejections, although no control over the build-up time of the acceleration is possible and more sudden onsets must be expected.

For negative (tailward) acceleration (downward ejection) no firm point for application of the accelerating force is accessible as for positive acceleration. If the force is applied as usual through harness and belt at shoulder and groin (87), the mobility of the shoulder girdle together with the elasticity of the belts results in a lower resonant frequency than the one observed in upward ejection. To avoid overshooting with standard harnesses the acceleration rise time must be at least 0.15 sec. It is obvious that this type of impact is able to excite the thorax-abdomen system (fig. 14); the diaphragm is pushed upwards by the abdominal viscera and as a result air rushes out of the lungs (if the glottis is open) or high pressures develop in the air passages (88). Tolerance limits for negative acceleration are probably set by the compression load on the thoracic vertebrae, which are exposed to the load of the portion of the body below the chest. This load on the vertebrae is more than for the positive acceleration case due to the greater weight; therefore a tolerance limit has been set at 13g. Shoulder accelerations of 13g have been tolerated by human subjects without injury, when the load was divided between hips and shoulders.

Transverse accelerations. - The forward-facing and backward-facing seated positions are most frequently exposed to high transverse components of crash loads. Human tolerance to these forces has been studied extensively by volunteer tests on linear decelerators (9), in automobile crashes (15) and by the analysis of the records of accidental falls (89). The results indicate the importance of distributing the decelerative

forces or impact over as wide an area as possible. The tolerable levels of well over 50g (100g and over for falling flat on the back with minor injuries, 35 to 40g for 0.05 sec. voluntary tolerance seated with restraining harness) are probably limited by injury to the brain. As indication that the latter might be sensitive to and based on specific dynamic responses is the fact that the tolerance limit depends strongly on the rise time of the acceleration. With rise times around 0.1 sec. (rate of change of acceleration 500 g/sec.) no overshooting of head and chest accelerations is observed, whereas faster rise times of around 0.03 sec. (1000 to 1400 g/sec.) results in overshooting of chest accelerations of 30 percent (acceleration front to back) and even up to 70 percent (acceleration back to front). All these results depend critically on the harness for fixation and the back support used (9). These dynamic load factors indicate a natural period of the body system between 10 and 20 c.p.s. No detailed physical studies of these parameters and of the importance of the thorax-abdomen system are available. Impact of the heart against the chest wall is another possible injury discussed and noted in some animal experiments (85,90).

The head and neck supporting structures seem to be relatively tough (9,15). Injury seems to occur only upon backward flexion and extension of the neck ("Whiplash") when the body is accelerated from back to front without head support as is common in rear-end automobile collisions.

Head impact. - Injuries to the head, beyond superficial bruises and lacerations, usually consist of concussion or fracture of the skull. The symptoms produced by head impact range from pain and dizziness through disorientation and depression of function to unconsciousness and loss of memory for events immediately preceding the injury (79). Head injuries usually occur from heavy blows by solid objects to the head, rather than by accelerative forces applied to the body. Since approximately 75 percent of airplane crash fatalities have been found to result from head injuries and the latter are without doubt of similar importance in many other types of accident (athletics, etc.) the mechanisms leading to head injury have been the object of a large number of investigations (23,86,91). In spite of this there is still considerable controversy over the physical mechanisms leading to injury and most of the information applies to specific impact situations. Thus no generalized, quantitative picture of the mechanical reaction of the head to the impact forces can be given at present.

Neck response. - The limitations for forward and lateral bending of the neck are, for practical purposes, the anterior chest wall and the shoulders respectively. Since the head is almost entirely held by the neck muscles, the absence of their supporting action, gives any blow to the head or neck a "flying start." As a consequence fractures or dislocations are more apt to result. Dislocations involving the first and second vertebral joints are usually less severe if the odontoid process is fractured (less damage to the spinal cord). If this is not the case the spinal cord may be severed or crushed. The latter is the essential mechanism in the hanging of criminals. The energy release required for this is around 1000 foot-pounds (150 kg. meters). As animal experiments indicate, damage done to the cervical cord at the first vertebra may also play a role in the generation of concussion, which has so far been attributed primarily to brain damage (77).

Head response. - The elastic shell of the skull is filled with nerve tissue, blood and cerebrospinal fluid, which have about the same density. The compressibility of the brain substance is very small (like water), and its shear modulus is very low. The viscosity of the brain tissue is around 20 dyne sec/cm ^{2}. The reaction of the head to a blow is a function of the velocity, duration and area of impact and the transfer of momentum. Near the point of application of the blow there will be an indentation of the skull. This results in shear strains in the brain in a superficial region close to the dent. Compression waves emanate from this area, which have normally small amplitudes since the brain is nearly incompressible. In addition to the forces on the brain resulting from skull deformation there are acceleration forces, which would also act on a completely undeformable skull. The centrifugal forces and linear accelerations producing compressional strains are negligible compared with the shear strains produced by the rotational accelerations. The distribution of the shear strains over the brain has been studied on models (92) and the motion of the brain surface has been observed in animals with sections of skull replaced by lucite (93). The maximum strains are concentrated at regions where the skull has a good grip on the brain owing to inwardly projecting ridges, especially at the wing of the sphenoid bone of the skull. Shear strains must also be present throughout the brain and in the brain stem. Many investigators consider these shear strains resulting from rotational

accelerations due to a blow to the unsupported head as the principal event leading to concussion. Blows to the supported, fixed head are supposed to produce concussion by compression of the skull and elevation of cerebrospinal fluid pressure. Despite the general acceptance of rotational acceleration as the main cause of concussion, experimental data on this quantity are almost completely missing and concussion thresholds are discussed in terms of "available energy" (which is usually not the energy transferred to the head) and impact velocity.

In general it can be assumed that a high velocity projectile (for example a bullet of 10 grams with a speed of 1000 ft/sec.) with its high kinetic energy and low momentum produces plainly visible injury to scalp, skull and brain along its path. The high frequency content of the impact is apt to produce compression waves which in the case of very high energies may conceivably lead to cavitation with resulting disruption of tissue (94). Skull fracture is not a prerequisite for these compression waves. On the other hand, if the head hits a wall or other object with masses large compared with the head's mass the local, visible damage is small and the invisible concussion damage due to rotational acceleration may be large. It must be added that blows to certain points, especially on the midline, produce no rotation. Blows to the chin upwards and sidewards on the other hand produce rotation relatively easily (knock-out in boxing). It is, therefore, almost impossible to define a concussion velocity or energy. Velocities listed in the literature for concussion from impact of large masses range from 15 to 50 ft/sec. At impact velocities of approximately 30 ft/sec. around 200 in.-lbs of energy was absorbed in 0.002 sec. resulting in an acceleration of the head of 47g. Impact energies for compression concussion are probably in the same range.

Skull response and skull fracture (95). - Impact studies on cadaver skulls with strain gauges show that a hammer blow to the bone itself lasts 2.5 to 5×10^{-4} sec. After the initial impact the bone oscillates for 2 to 4×10^{-3} sec. with a frequency of approximately 700 c.p.s. which agrees with the fundamental frequency found with continuous periodic excitation (43). Scalp, skin and subcutaneous tissue reduce the energy applied to the bone. If the response of the skull to a blow exceeds the elastic deformation limit, skull fracture occurs. Impact by a high velocity, blunt-shaped object results in localized circumscribed fracture and depression. Low velocity blunt blows insufficient to cause depression, occur frequently in falls and crashes. Stresscoat studies of such situations have revealed a general deformation pattern (95); an inward bent area of the skull surrounds the contact area and is followed at a considerable distance by an outbending of the skull. Sometimes fairly symmetrical undulating patterns are observed, occasionally with an outbending of the contrecoup type, i.e., an outbending in the skull area opposite the point of contact. Since the bone, a brittle material, fails in tension, most linear skull fractures originate in the outer surface of the outward bent area surrounding the indentation. From this area the crack may spread towards the center of impact, which rebounds immediately after the blow and becomes an area of high tensile stress. Propagation in the direction,opposite to the center of the impact takes place also. The size of the fracture line depends on the energy expended. Given enough energy two, three, or more cracks appear all radiating from the center of the blow. The skull has both weak and strong areas, each impact area showing well defined regions for the occurrence of the fracture lines.

The total energy required for skull fracture varies from 400 to 900 in.-lbs. with an average often assumed to be 600 in.-lbs. This energy is equivalent to the condition that the head hits a hard flat surface after a free fall from 5 ft. height. Skull fractures occurring when a batter is accidentally hit by a ball (5 oz.) of high velocity (100 ft/sec.) point towards the same energy (580 in.-lbs). Additional energy 10 to 20 percent beyond the single linear fracture demolishes the skull completely. Dry skull preparations required only approximately 25 in.-lbs for fracture. The reason for the large energy difference is mainly attributed to the attenuating properties of the scalp.

For practical situations in automobile and aircraft crashes the form, elasticity and plasticity of the object injuring the head is of extreme importance and determines its "head injury potential." For example, impact with a 90° sharp corner requires only a tenth of the energy for skull fracture (60 in.-lbs) that impact with a hard flat surface requires (96). Head impact energies for various attitudes of the human body hitting a contact area at various angles are presented in figure 24. The conditions shown are representative of crash conditions involving unrestrained humans.

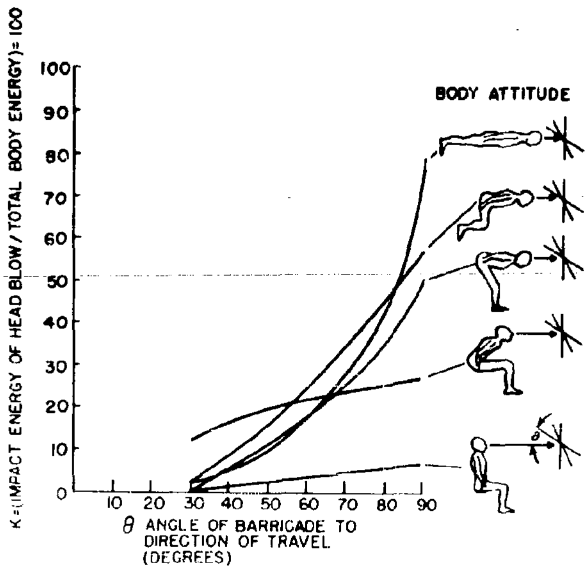

Figure 24. - Head impact energy as affected by body attitude and barricade angle [data from dummy drop tests after Dye (96)].

EFFECTS OF SHOCK AND VIBRATION ON TASK PERFORMANCE

As indicated above, very little is known about how shock or vibration affects task performance at accelerations which do not produce actual damage. One important factor is fatigue, but fatigue is not yet well enough understood to permit adequate measurement nor is it now possible to relate it quantitatively to any particular stress. Measurements of the performance of very simple tasks (36,97) have given equivocal or negative results except where there is mechanical interference. Experiments relating to task performance and fatigue are extraordinarily difficult to carry out properly and the results, when they are at all meaningful, have very limited applicability. "Quickie" experiments are almost invariably misleading.

PROTECTION METHODS AND PROCEDURES

Protection of man against mechanical forces is accomplished in two ways: isolation to reduce transmission of the forces to the man and increase of man's mechanical resistance to the forces. Isolation against shock and vibration is achieved by the standard suspension principle of having the natural frequency of the system to be isolated lower than the exciting frequency at least by a factor of 2. Both linear and nonlinear resistive elements are used for damping the transmission system; irreversible resistive elements or energy absorbing devices can be used once to change the time and amplitude pattern of impulsive forces. Human tolerance to mechanical forces is strongly influenced by selecting the proper body position with respect to the direction of forces to be expected. Man's resistance to mechanical forces can also be increased by proper distribution of the forces so that relative displacement of parts of the body is avoided as much as possible. This may be achieved by supporting the body over as wide an area as possible, preferably loading bony regions and thus making use of the rigidity available in the skeleton. Re-enforcement of the skeleton is an important feature of seats designed to protect against crash loads. The flexibility of the body is reduced by fixation to the rigid seat structure. The mobility of various parts of the body, e.g., the abdominal mass, can be reduced by properly designed belts and suits. The factor of training and indoctrination should also be mentioned as essential for the best use of protective equipment, for aligning the body into the least dangerous positions during intense vibration or crash exposure, and possibly for improving operator performance during vibration exposure. The latter type of training may be helpful in anticipating and preventing man-machine resonance effects and in reducing anxiety which might otherwise occur.

PROTECTION AGAINST VIBRATIONS

The transmission of vibration from a vehicle or platform to a man is reduced by mounting him on a spring or similar device, such as an elastic cushion (7,35). The degree of vibration isolation theoretically possible is limited in the important resonance frequency range of the sitting man by the fact that large static deflections of the man with the seat or into the seat cushion are undesirable and large relative movements between operator and vehicle control interfere in many situations with man's performance. A compromise is the only possible approach. Cushions are used primarily for static comfort but are also effective in decreasing transmission of vibration above man's resonance range. They are ineffective in the resonance range and may even amplify the vibration in the sub-resonance range. In order to achieve effective isolation over the 2 to 5 c.p.s. range, the natural frequency of the man-cushion system must be reduced to 1 c.p.s. but then the man would have to deflect into the cushion by 10 in. (fig. 25). Example of impedance changes achieved with various cushions are given in figure 26 (31). A condition known as "back scrub" may be produced as a result of using a seat cushion without a back cushion as is common in some tractor

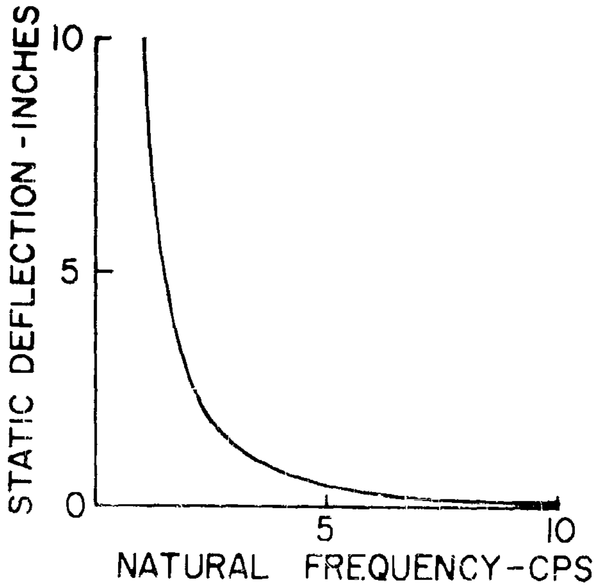

Figure 25. - Static deflection of a seat spring by the mass of the man as a function of the resonance frequency of the linear system. To tune the man-seat system lower than the resonance frequency of the man alone (5 c.p.s.) requires large static deflections.

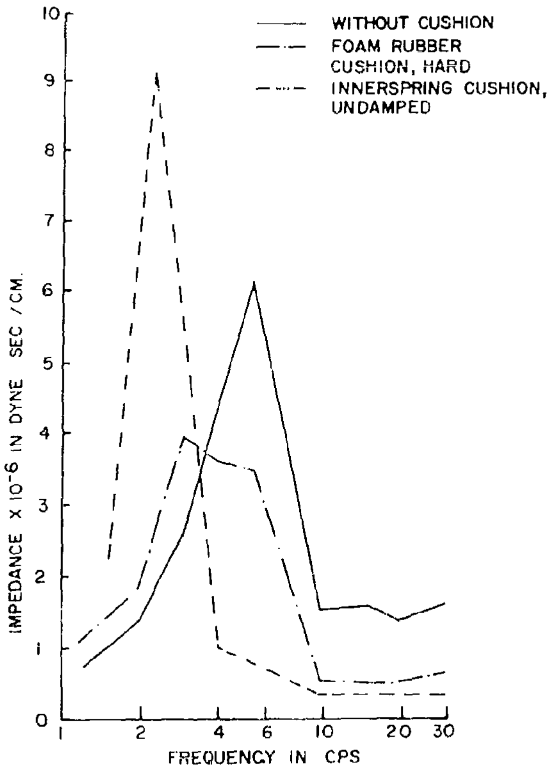

Figure 26. - Mechanical impedance of a sitting human subject with various cushions and without cushion (vertical, i.e., longitudinal vibrations) [from Dieckmann (31)].

or vehicle arrangements. Efforts of the operator to wedge himself between the controls and the back of the seat often tend to accentuate this uncomfortable condition. In ordinary passenger seats the seat cushion does not alter the resonance of the man-seat system considerably, i.e., in the low frequency range (< 5 c.p.s.) no isolation is achieved. Sometimes some amplification is unavoidable. Nevertheless the damping properties of the seat cushion are very important for attenuating the frequencies above resonance, which may be the more important areas, for example, in automobile or aircraft transportation (see fig. 33). For severe low frequency vibration and shock conditions, as they occur in tractors and other field equipment, suspension of the whole seat is therefore superior to the simple seat cushion. Hydraulic shock absorbers, rubber torsion bars, coil springs, and leaf springs have all been successfully used for suspension seats. The differences between these springs appear to be small. On the other hand there seems to be a difference in comfort and vibration induced stress if the seat is guided so that it can move only in a linear direction compared with the condition in which the seat simply pivots around a center of rotation (fig. 27) (7).

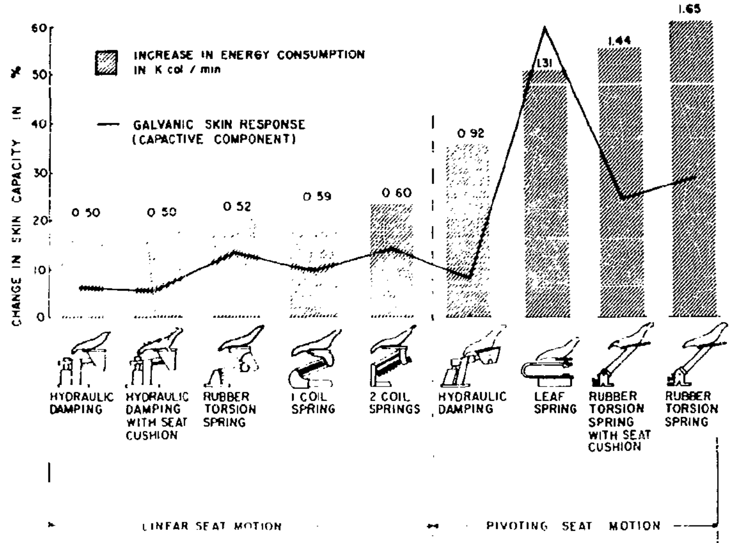

Figure 27. - Difference in vibration induced stress on a subject as a function of the seat design. Galvanic skin response and the increase in caloric energy consumption due to vibrations are plotted for the different seats indicated. The data represent averages over 10 tests of 15 minutes duration on five subjects with vibrations characteristic of tractor operation. The data prove the general superiority of seat designs which restrict seat vibrations to linear, compared with pivoting, motions. Subjective evaluations indicate a similar rank order as the physiological quantities in this graph [after Dieckmann (7)].

The latter situation produces uncomfortable and fatiguing pitching motion. Suspension seats can be built capable of preloading for the operator's weight to keep the static position of the seat and the natural frequency of the system at the desired value. Suspension seats for use on tractors and for similar applications are available which reduce the resonance of the man-seat system from approximately 4 to 1 c.p.s. (35) (measured in terms of the ratio of the acceleration at the belt level of the subject to the seat acceleration). The transmission ratio, which is between 2 and 2.5 at the resonance frequency for the subject alone, is only 1.6 at 1 c.p.s. for the subject on the suspension seat. With different types of foam and spring cushions alone it has not been possible to lower the resonance below 2 to 3 c.p.s. with a transmission ratio of 2.5 to 3. Elastic cushions alone result therefore in an amplification of the vibration in accordance with the impedance measurements (fig. 26). These findings are confirmed not only by laboratory tests but also by field tests on tractors. The superiority of man's legs to most seating devices with respect to the transmission of vertical vibrations has been shown in figure 9 and must be mentioned here for completeness. Differences in positioning of the sitting subject also change the transmission as demonstrated by impedance measurements (fig. 8). For severe vibrations close to or exceeding normal tolerance limits as they occur in military operations, special seats and restraints can be developed to provide maximum body support in all critical directions for the subject in the most advantageous position. In general, under these conditions, seat and restraint requirements are the same for vibration and rapidly applied accelerations. For a discussion of body restraints see below. The large protection to be expected by rigid or semi-rigid body enclosures has been proven by laboratory experiments (53) but has not yet been applied to practical situations. Immersion of the operator in a rigid, water-filled container with proper breathing provisions has recently been used to protect subjects against high, sustained static g-loads (98). It may be that the same principle could be used to provide protection against high alternating loads.

The isolation of hand and arm against vibration from hand tools depends critically on the type, weight, and relative position of the tool. Generalized rules are therefore hard to provide. The model for the hand-arm system presented in figure 16 should be adequate to estimate the effectiveness of isolation measures.

Since tolerance to continuous vibration depends critically on the exposure time, the control of working hours in vibration environments or with vibrating tools is one of the most important protective measures. It can prevent cumulative permanent damage and reduce the possibility of accidents favored by vibration-aggravated fatigue.

PROTECTION AGAINST RAPIDLY APPLIED ACCELERATIONS (CRASHES)

The study of automobile and aircraft crashes and of experiments with dummies and live subjects shows that complete body support and restraint of the extremities provide maximum protection against accelerating forces and give the best chance for survival (9,51). If the subject is restrained to the seat he makes full use of the force moderation provided by the collapsing of the vehicle structure and is protected against shifts in position which would injure him by bringing him in contact with interior surfaces of the cabin structure (99,100,101). The decelerative load must be distributed over as wide a body area as possible to avoid force concentration with resulting bending movements and shearing effects. The load should be transmitted as directly as possible to the skeleton, preferably directly to the pelvic structure and not via the vertebral column.

Theoretically a rigid envelope around the body would protect it to the maximum possible extent by preventing deformation. A body restrained to a rigid seat approximates such a condition; proper restraints against longitudinal acceleration shift part of the load of the shoulder girdle and arms from the spinal column to the back rest. Arm rests can take the load of the arms away from the shoulders (51). Semi-rigid and elastic abdominal supports provide some protection against large abdominal displacements. The effectiveness of this principle has been shown by animal experiments (38) and by impedance measurements on human subjects (fig. 8) but has not so far been applied in practice. Animals immersed in water, which distributes the load applied to the rigid container evenly over the body surface, or in rigid casts, have survived acceleration loads many times their normal tolerance.

Many attempts have been made to incorporate energy absorbing devices either in a harness or in a seat with the intent to change the acceleration time pattern by limiting peak accelerations. Parts of the seat or harness are designed with a nonlinear characteristic which starts to extend at some given acceleration level. Unfortunately the benefits derived from such a device are usually small since little space for body- or seat-motion is available in airplanes or automobiles and contact with interior cabin surfaces during the period of extension of the device is apt to result in more serious injury. For example, (fig. 23) an aircraft

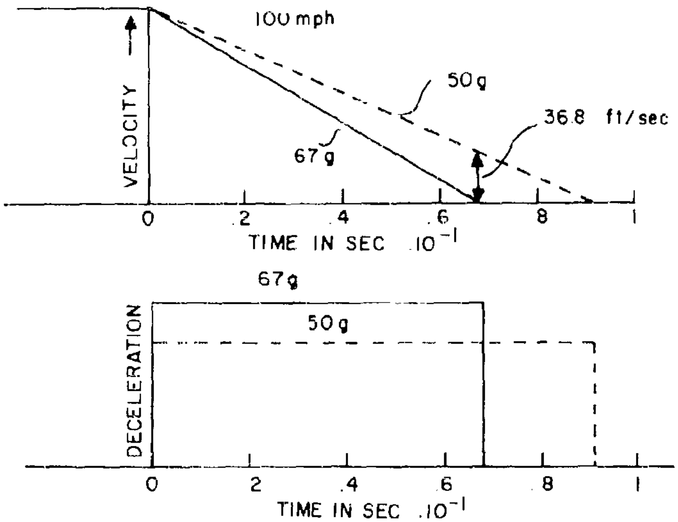

Figure 28. - Crash deceleration and velocity of aircraft as a function of time. The aircraft comes to a complete stop from 100 m.p.h. within 5 feet. An energy absorbing device limiting the maximum acceleration on the passenger to 50g is assumed. It would require a displacement of 19 in.

being stopped in a crash from 100 m.p.h. in 5 ft. would be subjected to a constant deceleration of 64.4g. An energy absorbing device designed to elongate at 50g would require 19 in. displacement. While traveling this distance the body or seat would be decelerated relative to the aircraft by 14.4g and would have a maximum velocity of 36.8 ft/sec. relative to the aircraft structure. A head striking a solid surface with this velocity would have many times the minimum energy needed for a fractured skull. The available space for seat or passenger travel using the principle of energy absorbtion must therefore be considered carefully in the design. In the development of catch-up mechanisms for window washers and workers in similar situations energy absorbing elements in the form of undrawn nylon ropes may have promising applications. For seat belts and other crash restraint harnesses extensible fabrics have so far been found to be extremely hazardous since their load characteristics cannot be sufficiently controlled (102). Recently seats for jet airliners have been designed which have extendable rear legs housing energy absorbing mechanisms (103). The maximum travel of the seats is 6 in. and their motion is designed to start between 9 and 12g horizontal load depending on the floor strength. During motion the legs pivot at the floor level, a feature also supposed to be beneficial if the floor wrinkles in the crash. Theoretically such a seat could be exposed to 30g for 0.037 seconds or 20g for 0.067 seconds and have not more than 9g transmitted to the seat. Full test reports on these seats are not available.

The high tolerance limits of the well supported human body to decelerative forces (fig. 34 to 41) suggest that in aircraft and other vehicles, seats, floors, and the whole inner structure surrounding crew and passengers should be designed to resist crash forces as near to 40g as weight or space limitations permit (103). The structural members surrounding this inner compartment should be arranged so that their crushing reduces forces on the inner structure. Protruding and easily loosened objects should be avoided. To allow the best chance for survival, seats should also be stressed for dynamic loadings between 20 and 40g. Civil Air Regulations require 9g static strength of seats as a minimum (104). A method for computing seat tolerance to typical survivable airplane crash decelerations is available for seats of conventional design (105). There is no question that the passenger who is riding in a seat facing backward has a better chance to survive an abrupt crash deceleration since the impact forces are then more uniformly distributed over the body. Neck injury has to be prevented by proper head support. The objections to riding backwards on a railway or in a bus are not present for air transportation because of the absence of disturbing motion of objects in the immediate field of view. The controversy over the question whether passenger seats in aircraft should face forward or aft stems from the fact that for a rearward facing seat the center of passenger support is about 1 ft. above the point where the seat belt would be attached for a forward facing passenger. Consequently the rearward facing seat is subjected to a higher bending moment; in other words, for seats of the same weight the forward facing seat will sustain higher crash forces without collapse. For the same seat weight the rearward facing seat sould have approximately only half the design strength of the forward facing seat and about one-third its natural frequency. A criterion for the selection of one type seat over the other on the basis of allowable weight, seat tolerance to probable crash loads, and passenger injury has recently been proposed but must await experiments for quantitative confirmation (106).

Safety lap or seat belts are used to fix the occupants of aircraft or automobiles to their seats and to prevent their being hurled about within the car or aircraft or being ejected from the car. Their effectiveness

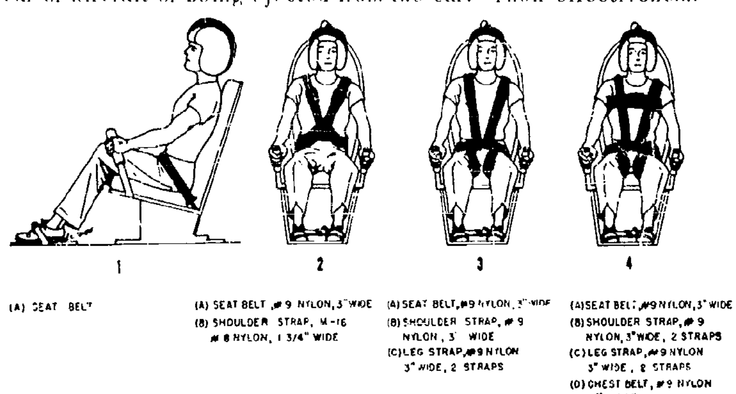

Figure 29. - Protective harness arrangements for rapid accelerations or decelerations.
1. Seat belt for automobiles and commercial aviation.
2. Standard military lap and shoulder strap.
3. Like 2 but with thigh straps added to prevent headward rotation of the lap strap.
4. Like 3 but with chest strap added. These arrangements were evaluated in sled deceleration tests [from Stapp (9)].

Table 4. Human-Body Restraint and Possible Increased Impact Survivability [After Eiband (51)]

Direction of Acceleration Imposed on Seated Occupants	*Conventional Restraint*	*Possible Survivability Increases Available by Additional Body Supports**
Spineward: Crew	Lap strap Shoulder strap	Forward facing: (a) Thigh straps (Assume crew members will be performing emergency duties with hands and feet at impact.)
Passengers	Lap strap	Forward facing: (a) Shoulder straps (b) Thigh straps (c) Nonfailing arm rests (d) Suitable hand-holds (e) Emergency toe straps in floor
Sternumward: Passengers only	Lap strap	Backward facing: (a) Nondeflecting seat back and (b) Integral, full-height head rest (c) Chest strap (axillary level) (d) Lateral head motion restricted by padded "winged back" (e) Leg and foot barriers (f) Arm rests and hand-holds (prevent arm displacement beyond seat back)
Headward: Crew	Lap strap Shoulder straps	Forward facing: (a) Thigh straps (b) Chest strap (axillary level) (c) Full, integral head rest (Assume crew members will be performing emergency duties; extremity restraint useless.)
Passengers	Lap strap	Forward facing: (a) Shoulder straps (b) Thigh straps (c) Chest strap (axillary level) (d) Full, integral head rest (e) Nonfailing contoured arm rests (f) Suitable hand-holds Backward facing: (a) Chest strap (axillary level) (b) Full, integral head rest (c) Nonfailing arm rests (d) Suitable hand-holds
Tailward: Crew	Lap strap Shoulder straps	Forward facing: (a) Lap-belt tie-down strap (Assume crew members will be performing emergency duties; extremity restraint useless.)
Passengers	Lap strap	Forward facing: (a) Shoulder straps (b) Lap-belt tie-down strap (c) Hand-holds (d) Emergency toe straps Backward facing: (a) Chest straps (axillary level) (b) Hand-holds (c) Emergency toe straps
Berthed Occupants	Lap strap	Feet forward: Full-support webbing net Athwart ships: Full-support webbing net

* Exposure to maximum tolerance limits (figs. 34 to 41) requires straps exceeding conventional strap strength and width.

has been proven by many laboratory tests and in actual crash accidents (86). The belt load on the lower abdomen causes no severe intra-abdominal injury or injury to the lower spinal region at least in most survivable crashes. A forward facing passenger held by a seat belt flails about when suddenly decelerated; his hands, feet, and upper torso swing forward until his chest hits his knees or until the body is stopped in this motion by hitting other objects (back of seat in front, cabin wall, instrument panel, steering wheel, control stick, etc.). Since 15 to 18g longitudinal deceleration can result in three times higher acceleration of the chest hitting the knees, this load appears to be about the limit a human can tolerate with a seat belt alone. Approximately the same limit is obtained when the head-neck structure is considered. Increased safety in automobile as well as airplane crashes could be obtained by distributing the impact load over larger areas of the body and fixing the body more rigidly to the seat. Shoulder straps, thigh straps, chest straps, and hand holds are additional body supports used in experiments. They are illustrated in figure 29. Table 4 shows the desirability of these additional restraints to increase possible survivability to acceleration loads of various direction (51). In airplane crashes vertical and horizontal loads must be anticipated. In automobile crashes horizontal loads are most likely and tests should verify the lack of merit in providing much more than adequately engineered shoulder or chest straps. The effectiveness of these configurations in automobile crashes is illustrated in figure 30. Lap straps should always be as tight as comfort will permit to exclude slack as far as possible. During forward movement 60 percent of the body mass is restrained by the belt, i.e., must be considered as belt load. Instead of the 3000 to 4000 lb., three-inch lap belt currently in wide use an 8000 lb., loop strength, three-inch wide nylon belt is recommended to reduce slack and elongation under load (100). Double thickness number 9 undrawn nylon straps of 3 in. width have been found most satisfactory for all harnesses (50) with respect to strength, elongation, and supported surface area. If the upper torso is fixed to the back of the seat by any type of harness (shoulder harness, chest belt, etc.), the load on the seat is approximately the same for forward and backward facing seats. The difference between these seats with respect to crash tolerance as discussed above no longer exists. Research to simplify and adapt these body restraints for passenger and crew use without creating too much discomfort is continuing (100,107). So far restraints in addition to lap belts are routinely only used in military aviation.

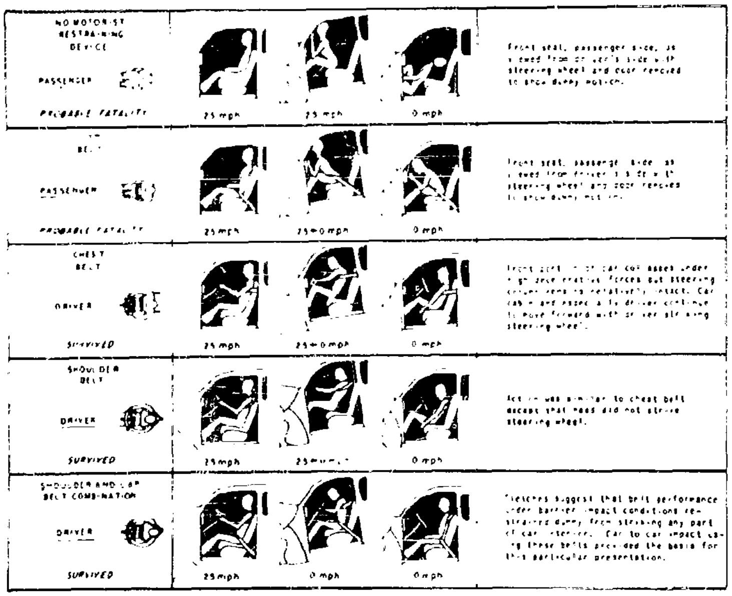

Figure 30. - Effect of varying safety-belt arrangements on driver and passenger for a 25 m.p.h. automobile collision with a fixed barrier. The sketches and evaluations are based on actual collision tests [from Severy and Mathewson (99)].

The dynamic properties of seat cushions are extremely important if an acceleration force is applied through the cushion to the body. In this case the steady state response curve of the total seat-man system (fig. 26) gives a clue to the possible dynamic load factors under impact. Overshooting should be avoided, at least for the most probable impact rise times. This problem has been studied in detail in connection with seat cushions used on upward ejection seats (87). The ideal cushion is approached when its compression under

static load spreads the load uniformly and comfortably over a wide area of the body and if almost full compression is reached under the normal weight. The impact acceleration then acts uniformly and almost directly on the body without intervening elastic elements. A slow responding foam plastic, for example, with thickness of 2 to 2.5 in. has been found to be very satisfactory in fulfilling these requirements.

Safety engineering for crash worthiness of automobiles and aircraft, their seats and restraints, has still a long way to go before maximum use can be made of the human body's deceleration tolerance limits.

PROTECTION AGAINST HEAD IMPACT

The impact reducing properties of protective helmets are based on two principles: the distribution of the load over a large area of the skull and the interposition of energy absorbing systems. The first principle is applied by using a hard shell, which is suspended by padding or a support webbing at some distance from the head (5/8 in. to 3/4 in.). High local impact forces are distributed by proper suports over the whole side of the skull to which the blow is applied. Skull injury from relatively small objects and projectiles can thus be avoided. However, tests usually show that contact padding alone over the skull results in most instances in greater load concentration, whereas helmets with web suspension distribute pressures uniformly (23). Since helmets with contact padding usually permit less slippage of the helmet, a combination of web or strap suspension with contact padding is desirable. The shell itself must be as stiff as is compatible with weight considerations and not be permitted to deflect sufficiently under blows to come in contact with the head. For light industrial safety hats molded bakelite reinforced with steel wire, laminated bakelite, or high-strength aluminum alloy are used; for football helmets combination rubber and plastic compounds molded in a single shell or a vulcanized fiber shell have been used; for anti-buffet helmets molded fiberglass flock, molded cotton fabric or cloth laminates, all impregnated with plastic are some of the most commonly used materials.

Padding materials can incorporate energy absorbing features. Whereas foam rubber and felt are too elastic to absorb a blow, foam plastics like polystyrene or Ensolite have been found to result in lower transmitted accelerations. Little is known about how well the objectives of energy absorption (reduction of peak acceleration and change in high rate of acceleration onset) have been realized in practical designs.

Since it is very hard to spell out the exact physical conditions for a helmet which can provide optimum impact protection, quantitative evaluation of the effectiveness of various helmet designs is difficult. In addition most of these helmets constitute compromises among several objectives such as pressurization, communication, temperature conditioning, minimum bulk and weight, visibility, protection against falling objects, etc. Impact protection is therefore usually only one of many design features and something like an "optimum design" for impact protection alone has not been worked out. The protective effect of helmets against concussion and skull fracture has been proven in animal experiments (108) and is apparent from accident statistics.

PROTECTION AGAINST BLAST WAVES

Individual protection against air blast waves is extremely difficult since only very thick protective covers can reduce the transmission of the blast energy significantly. Furthermore, not only the thorax but the whole trunk would probably have to be covered. In animal experiments sponge rubber wrappings and jackets of other elastic material have resulted in some reduction of blast injuries (5). Enclosure of the animal in a metal cylinder with only the head exposed to the blast wave has provided the best protection short of complete enclosure of the animal. It is therefore generally assumed that shelters are the only practical means of protecting humans against blast. They may be of either the open or closed type; both change the pressure environment. Changes in pressure rise time introduced by the door or other restricted openings are physiologically most important (17). Reflection phenomena within the shelter must also be considered.

Protection against water blast waves is obtained from air containing vests which partially reflect the blast waves. No quantitative information on this effect is available.

TOLERANCE CRITERIA FOR VARIOUS TYPES OF EXPOSURE AND ACTUAL ENVIRONMENTS EXPERIENCED BY MAN

The previous paragraphs should have made it clear how difficult it is to predict the effect of mechanical forces on man. Even if the experimental material available were more exact and complete, it would still be impossible to give exact limits for man's safety and performance under field conditions since the exact physical mode of action of the environment varies with man's unpredictable position and motion and since biological variations with respect to physical, physiological, and psychological reactions make such limits only a statistical quantity. Individual cases may deviate considerably from the average. Biological criteria must therefore be used with caution; they are summarized here only as rough guides for engineering purposes. Tolerance criteria as well as examples of field environment indicate only orders of magnitude and are not rigid limits. For critical safety problems, detailed study of the literature or expert consultation is indispensible.

VIBRATION EXPOSURE

In spite of the fact that many schemes have been developed to assess man's reaction to vibration in a quantitative manner most of them are based on a limited number, specific types, or a specific interpretation of experiments and contradict each other to a certain degree. The averaging of all these results and their simplification to the broad groups I to III in figure 31 seems therefore the most reasonable approach (72).

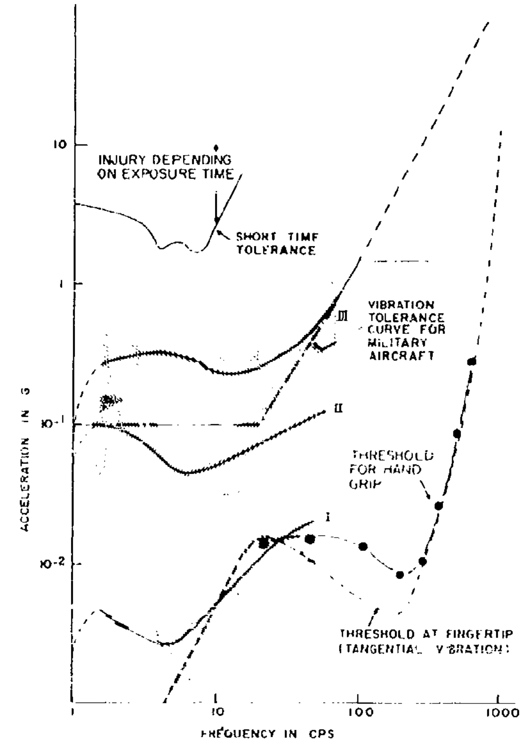

Figure 31. - Vibration tolerance criteria. Average peak accelerations at which subjects perceive vibrations, I; find it unpleasant, II; or refuse to tolerate it further, III. The shaded areas are one standard deviation on either side of the mean. These curves are for subjects without any protection, exposure times 5 to 20 minutes. The short time tolerance curve is for subjects with standard Air Force lap belt and shoulder harness, exposure time approximately one minute. The WADC vibration tolerance curve is used for long-time exposure in military aircraft. [After Goldman (72), Getline, (73), Ziegenruecker and Magid (74), von Békésy (20) and Fibikar (111).]

These criteria have been used widely to classify the severity of vibration exposure. They represent averages for the standing, sitting, and lying positions. For vibrations in several directions, the vector sum of all components is proposed as the acceleration stimulus to be used for evaluating the condition. Larger accelerations than those indicated by area III in the figure can be tolerated by the majority of young, male subjects for short time periods only without harmful effects. The curve marked "short time tolerance" has been established for exposure times in the order of one minute or less for young, male military subjects strapped in an airplane seat with seat belt and shoulder harness (74). This curve must be considered the borderline beyond which physical tissue damage occurs in relatively short time. The threshold for vibration reception at the fingertip (20) is also shown in the figure and might be helpful in explaining annoying vibration transmitted to parts of the body like the hands or the head.

For irregular, random vibrations the tolerance curves established for sinusoidal vibrations must also serve as guidelines for the time being, since no better data exist. Single shock acceleration pulses as they occur as floor vibrations near drop forges or similar equipment were used in one study and some of the results are presented in figure 32 (12). The intolerable range in this series of experiments should probably be considered as conservative since experiments with sinusoidal vibrations by the same authors, which were included in the averages of figure 31, gave relatively conservative tolerance levels.

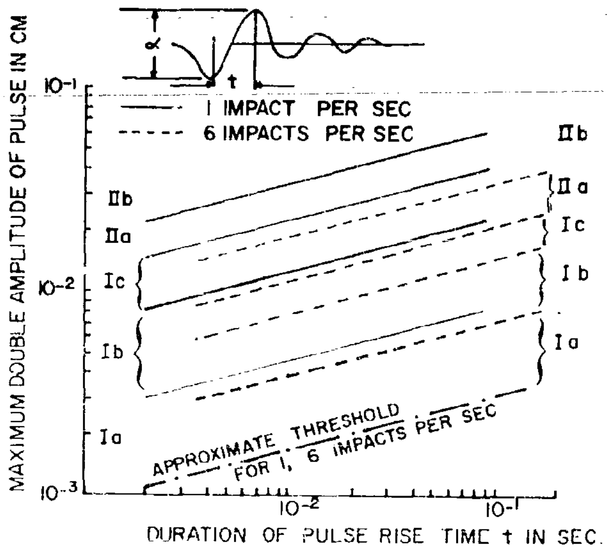

Figure 32. - Tolerance of human subjects in the standing or supine position to repetitive vertical impact pulses representative of impacts from pile drivers heavy tools, heavy traffic, etc. Subjective reaction is plotted as a function of the maximum displacement of the initial pulse and its rise time. The numbers indicate the following reactions for the areas between the lines: Ia, threshold of perception; Ib, of easy perception; Ic, of strong perception, annoying; IIa, very unpleasant, potential danger for long exposures; IIb, extremely unpleasant, definitely dangerous. The decay process of the impact pulses was found to be of little practical significance [after Reiher and Meister (12)].

The range of vibration levels found in aircraft, trucks, and tractors is indicated in figure 33. Individual points represent flight vibration data obtained in various types of military aircraft (73). The solid circles, squares, and triangles indicate vibrations at seat levels which were reported as excessive and undesirable in actual service; the open marks indicate conditions accepted as tolerable. The linearized dividing line between tolerable and excessive vibrations is the WADC tolerance criterion shown in figure 31, used as a

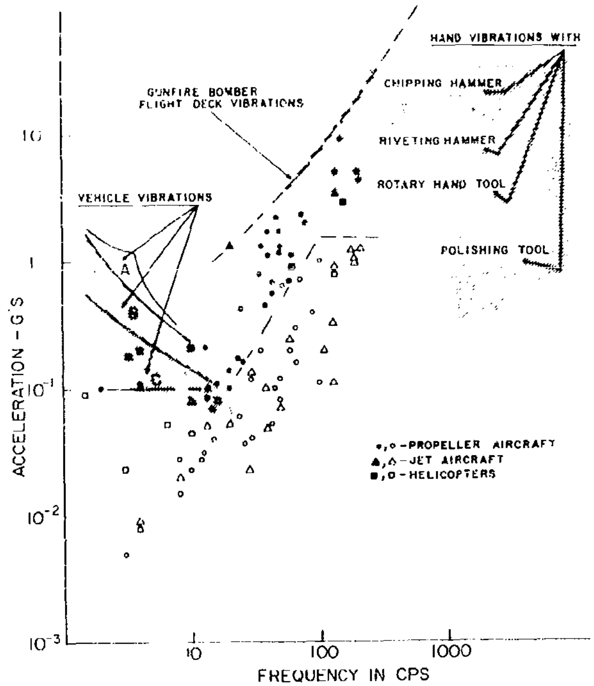

Figure 33. - Approximate vibration environments for automobiles, trucks, tractors, and aircraft. Vibration levels observed on hands while operating various hand tools are also indicated. These levels are strongly dependent on tool design and type of operation. Most of the hand vibrations indicated are in the acceleration range at which chronic hand injuries may result. [After Radke (35), Dieckmann (7), Getline (73), and Agate and Druett (41) and others.]

long time vibration tolerance criterion in military aviation. The areas indicated for truck and tractor vibration (7,35) are the range for the respective vibration maxima and do not represent spectra. Most vehicles have very pronounced natural frequencies excited according to ground conditions. Very generally, rubber-tired farm tractors, as well as trucks, have the maximum of their vertical accelerations in the 2 to 6 c.p.s. range; for large rubber-tired earth moving equipment the range is 1.5 to 3.5 c.p.s. and for crawler type tractors it is near 5 c.p.s.

DECELERATION EXPOSURE, CRASH, AND IMPACT

The approximate maximum tolerance limits to rapid decelerations applied to a sitting human subject are summarized in figures 34 to 41 (51). The data are compared and summarized on the basis of trapezoidal pulses on the seat in all four acceleration directions with respect to a saggital plane through the body axis.

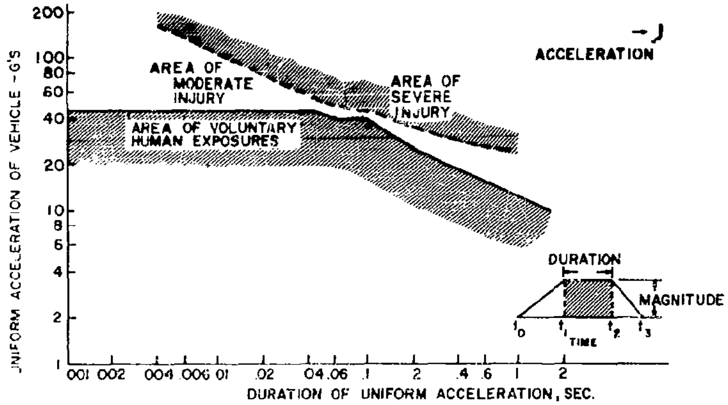

Figure 34. - Tolerance to spineward acceleration as a function of magnitude and duration of impulse [from Eiband (51)].

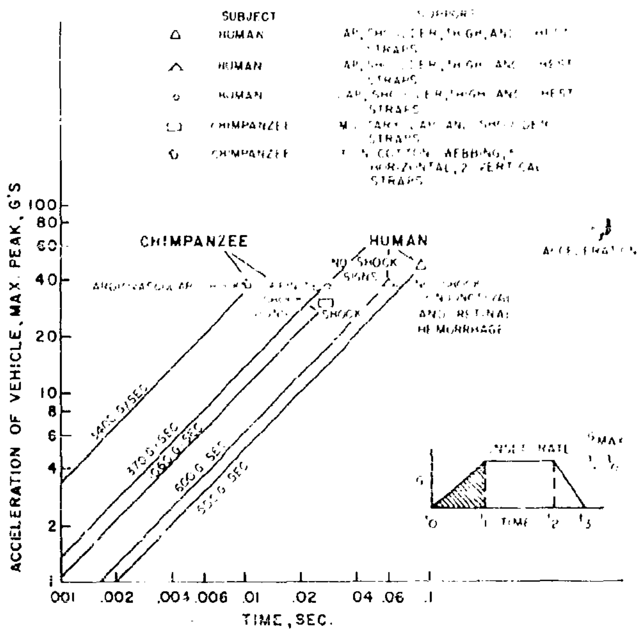

Figure 35. - Effect of rate of onset on spineward acceleration tolerance [from Eiband (51)].

The limits as shown are based on experiments providing maximum body support (see table 4), i.e., lap belt, shoulder harness, thigh and chest strap, and arm rests for the headward accelerations. The quantitative influence of the initial rate of change of acceleration is not too clearly established and not enough data are available for exact mathematical analysis of the influence of the total acceleration-time function. Although the separation of this function into duration of (uniform) acceleration and onset rate is not completely

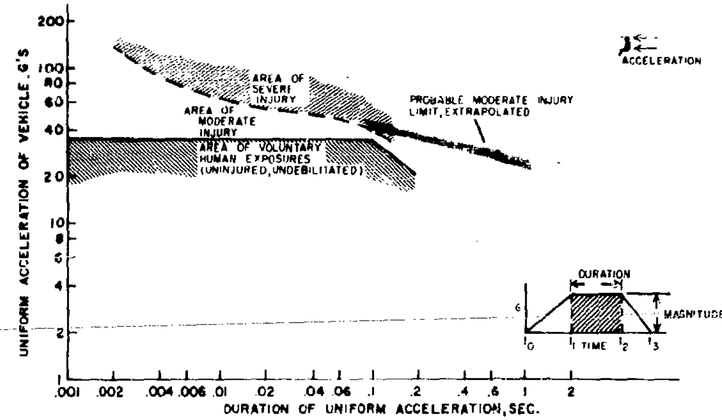

Figure 36. - Tolerance to sternumward acceleration as a function of magnitude and duration of impulse [from Eiband (51)].

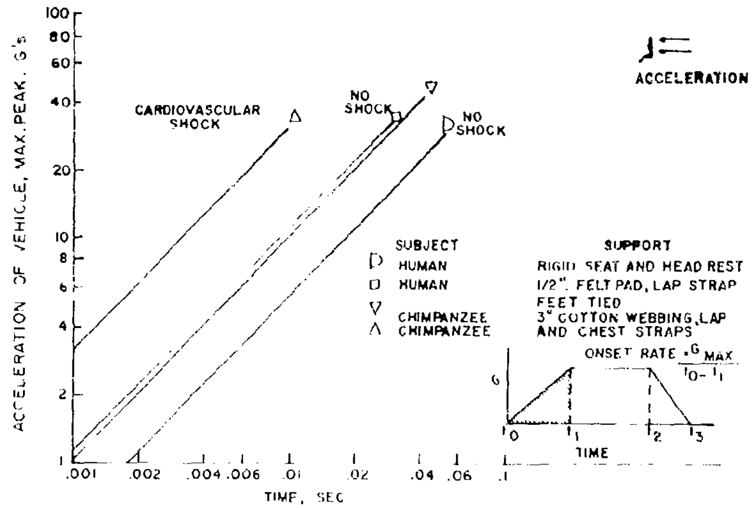

Figure 37. - Effect of rate of onset on spineward acceleration tolerance [from Eiband (51)].

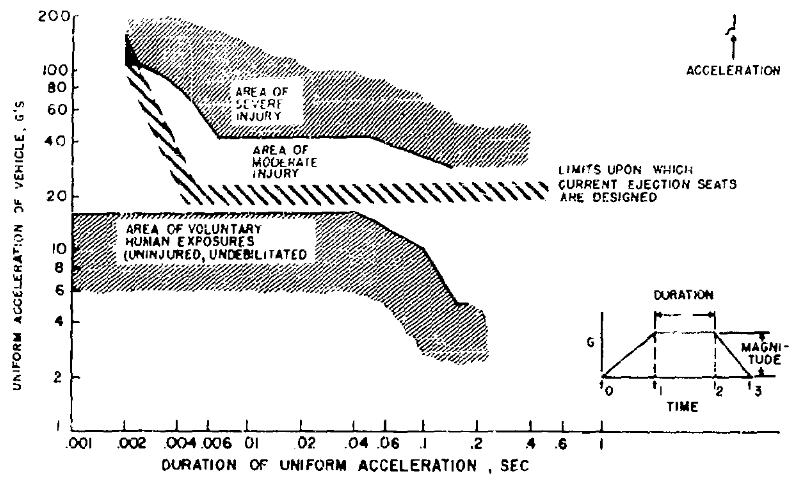

Figure 38. - Tolerance to headward acceleration as a function of magnitude and duration of impulse [from Eiband (51)].

satisfying, it constitutes the most useful analysis of the experimental evidence available. Onset rates endured by various subjects are therefore summarized in separate graphs for the different directions (figs. 35,37,39,41). Caution in applying the curves presented must again be stressed, since they are based on well designed body supports, minimum slack of the harnesses, heavy seat construction, young, healthy

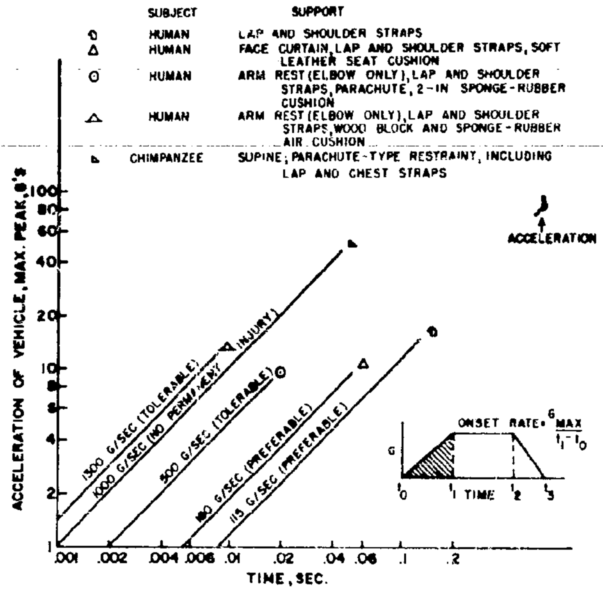

Figure 39. - Effect of rate of onset on headward acceleration tolerance [from Eiband (51)].

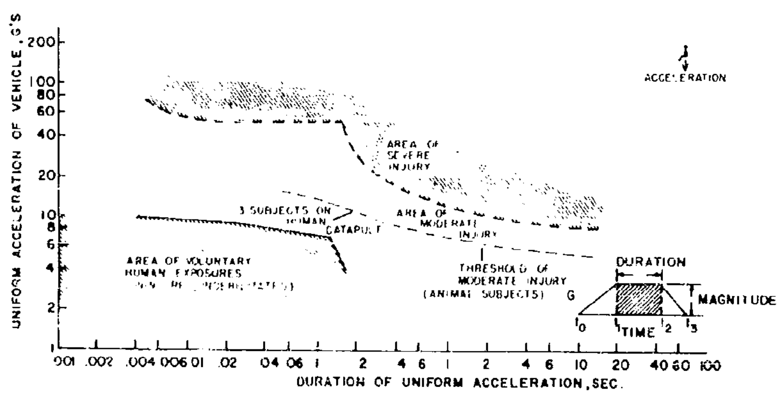

Figure 40. - Tolerance to tailward acceleration as a function of magnitude and duration of impulse [from Eiband (51)].

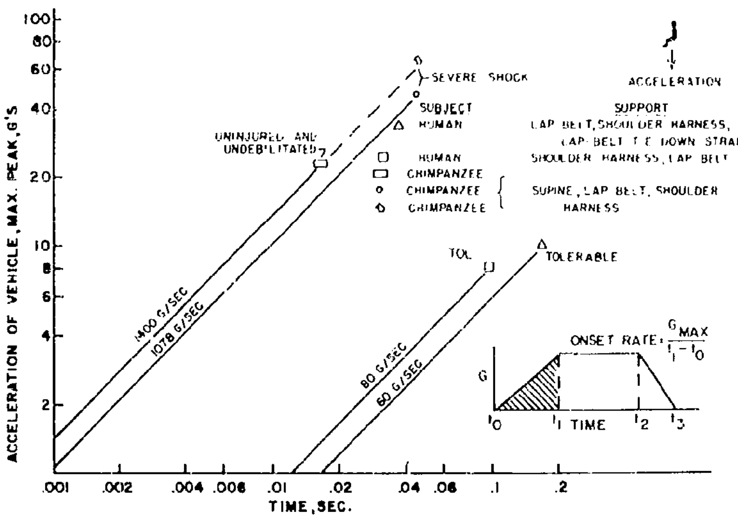

Figure 41. - Effect of rate of onset on tailward acceleration tolerance [from Eiband (51)].

volunteer subjects, and subjects expecting the impact exposure. Thus these curves constitute maxima in many respects, although further improvement in the protection methods does certainly not appear impossible. For example, the maximum limit for exposure to transverse front-to-back acceleration, as experienced in head-on automobile collisions, is indicated in figure 34 as between 40 and 50g for durations of less than 0.1 sec. For subjects without maximum upper torso restraint having only a lap belt or other types of abdominal restraints, this limit is estimated to be between 10 and 20g.

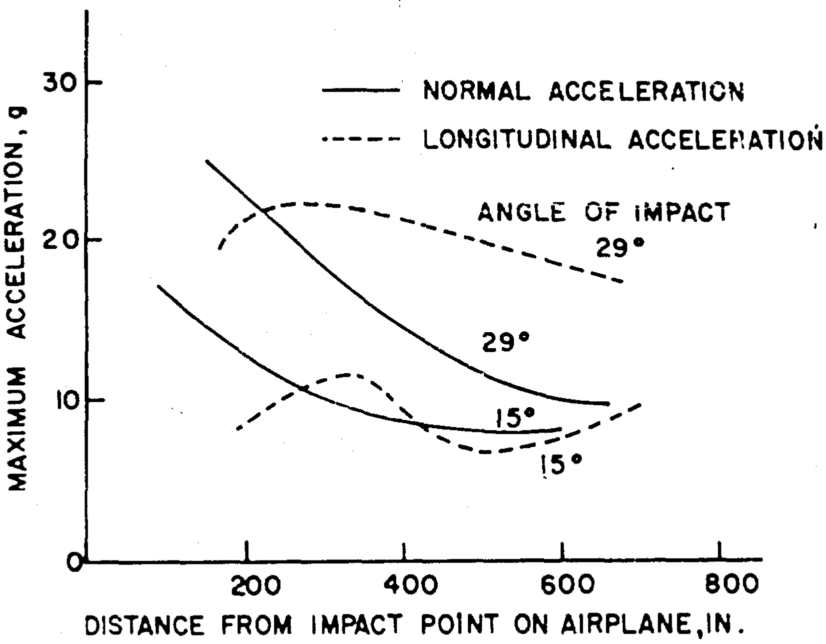

Figure 42. - Longitudinal and normal crash loads on a pressurized transport aircraft hitting the ground at 35 m.p.h. under an impact angle of 15° and 29°. Acceleration levels in the aircraft are shown as a function of the distance from the point of contact (nose) [after Preston and Pesman (109)].

Approximate ranges for aircraft crash loads can be obtained from figure 42 (109). Horizontal crash loads, i.e., in the direction of the aircraft's longitudinal axis, increase with the crash angle to a maximum at 90°, whereas vertical loads reach their maximum approximately at 35°. The graph shows only one typical example; aircraft type, ground conditions, and point of initial crash contact have a strong influence in each individual case. For automobile head-on collisions, figure 43 shows typical deceleration patterns for the

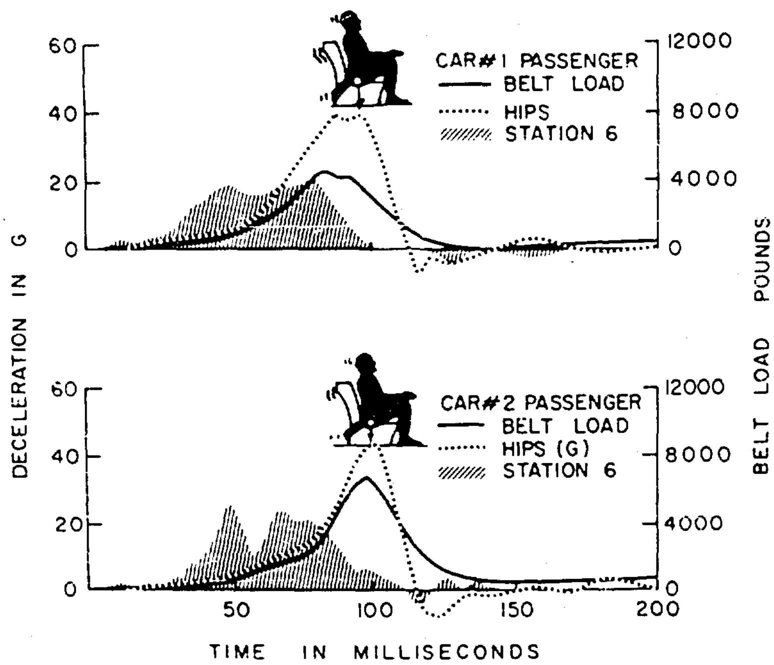

Figure 43. - Example of automobile head-on collision deceleration patterns as a function of time. The deceleration for the underbody under the seat and the passenger's hip is plotted together with the load function of the seat belt. The data are for two cars engaged in experimental head-on collisions. (Impact speed 31 ft/sec. Kinetic energy of cars before impact approximately 45,000 ft/lbs. Cars collapse under impact approximately 1.7 ft.) [From Severy *et al.* (100).]

car structure under the seat and the passenger's hips; seat belt loads are also indicated in the graph (100). The two graphs in this figure are for two cars colliding with each other head-on. Figure 44 summarized the results of many automobile crash experiments. The peak deceleration of the car body under the driver's compartment is plotted against the impact velocity. As can be seen from the graph, the difference in impact load between frame and unitized underbody construction was negligible in these experiments.

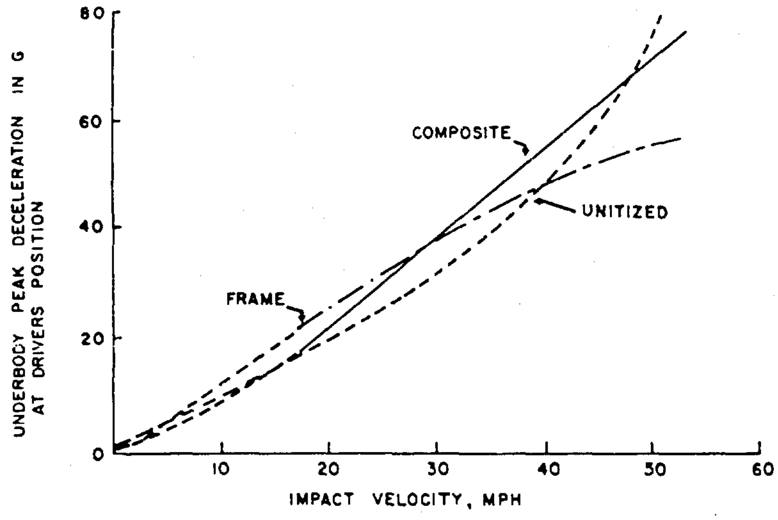

Figure 44. - Crash deceleration of the passenger compartment in head-on collisions of automobiles as a function of driving speed. The negligible difference between frame and unitized underbody construction is also shown. [From Severy *et al.* (100).]

Examples of other types of short duration accelerations are given in table 5.

Table 5. Approximate Duration and Magnitude of Some Short Duration Acceleration Loads (From various sources)

Type of Operation	Acceleration (g)	Duration (sec.)
Elevators: average in "fast service"	.1 - .2	1 - 5
comfort limit	.3	
emergency deceleration	2.5	
Public transit: normal acceleration and deceleration	.1 - .2	5
emergency stop braking from 70 m.p.h.	.4	2.5
Automobiles: comfortable stop	.25	5 - 8
very undesirable	.45	3 - 5
maximum obtainable	.7	3
crash (potentially survivable)	20 - 100	<.1
Aircraft: ordinary take-off	.5	>10
catapult take-off	2.5 - 6	1.5
crash landing (potentially survivable)	20 - 100	
seat ejection	10 - 15	.25
Man: parachute opening – 40,000 ft.	33	.2 - .5
6,000 ft.	8.5	.5
parachute landing	3 - 4	.1 - .2
fall into fireman's net	20	.1
approximate survival limit with well-distributed forces (fall into deep snow bank)	200	.015-.03
Head: adult head falling from 6 ft. onto hard surface	250	.007
voluntarily tolerated impact with protective headgear	18 - 23	.02

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